EVALUATION OF VON MISES STRESSES IN FOUR BONE QUALITIES IN AN IMPLANT SUPPORTED MANDIBULAR POSTERIOR CROWN BY 3D FINITE ELEMENT ANALYSIS

A Dissertation submitted to the

THE TAMILNADU DR. MGR MEDICAL UNIVERSITY

In partial fulfillment of the requirements
for the degree of

MASTER OF DENTAL SURGERY

(BRANCH – I)

(PROSTHODONTICS AND CROWN & BRIDGE)

2015 – 2018

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CERTIFICATE

This is to certify that **DR.V.P. KIRUTHIGA,,** Post Graduate student (2015 - 2018) in the Department of Prosthodontics and Crown and Bridge, has done this dissertation titled **"EVALUATION OF VON MISES STRESSES IN FOUR BONE QUALITIES IN AN IMPLANT SUPPORTED MANDIBULAR POSTERIOR CROWN BY 3D FINITE ELEMENT ANALYSIS"** under my direct guidance and supervision in partial fulfillment of the regulations laid down by **The Tamil Nadu Dr. M.G.R. Medical University, Guindy, Chennai – 32** for **M.D.S.** in **Prosthodontics and Crown & Bridge (Branch I)** Degree Examination.

Guided by

PROF.Dr.C.SABARIGIRINATHAN.M.D.S.
Head of the Department
Department of Prosthodontics,
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Head of the institution

PROF.Dr.B.SARAVANAN.M.D.S.,Ph.D.
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Tamil Nadu Govt Dental College
and Hospital, Chennai - 600 003.

DECLARATION

I, **Dr. V.P KIRUTHIGA** do hereby declare that the dissertation titled **" EVALUATION OF VON MISSES STRESSES IN FOUR BONE TYPES IN A IMPLANT SUPPORTED MANDIBULAR POSTERIOR CROWN BY 3D FINITE ELEMENT ANALYSIS"** was done in the Department of Prosthodontics, Tamil Nadu Government Dental College & Hospital, Chennai 600 003. I have utilized the facilities provided in the Government Dental College for the study in partial fulfilment of the requirements for the degree of **Master of Dental Surgery** in the speciality of **Prosthodontics and Crown & Bridge (Branch I)** during the course period **2015-2018** under the conceptualization and guidance of my dissertation guide, **PROF. Dr. C. SABARIGIRINATHAN.MDS.,** and my Co-guide, Associate **Prof. Dr. P. RUPKUMAR, M.D.S.**

I declare that no part of the dissertation will be utilized for gaining financial assistance for research or other promotions without obtaining prior permission from the Tamil Nadu Government Dental College & Hospital.

I also declare that no part of this work will be published either in the print or electronic media except with those who have been actively involved in this dissertation work and I firmly affirm that the right to preserve or publish this work rests solely with the prior permission of the Principal, Tamil Nadu Government Dental College & Hospital, Chennai 600 003, but with the vested right that I shall be cited as the author(s).

Signature of the PG student **Signature of the HOD** **Signature of the Co- Guide**

Signature of the Head of the Institution

TRIPARTITE AGREEMENT

This agreement herein after the "Agreement" is entered into on this day, ____________________ between the **Tamil Nadu Government Dental College and Hospital** represented by its Principal having address at Tamil Nadu Government Dental College and Hospital, Chennai-3, (hereafter referred to as, 'the College')

And

Dr. C. SABARIGIRINATHAN,M.D.S., aged 50 years working as Professor and Head of Department of Prosthodontics and crown and bridge at Tamil Nadu Government Dental College and Hospital, Chennai-3 having residence address at E/32, Anna Nagar (East), Chennai- 102 (herein after referred to as the 'Researcher and Principal investigator') and **Dr. P. RUPKUMAR, M.D.S.,** aged 45 years working as Associate Professor, Department of Prosthodontics and crown and bridge at Tamil Nadu Government Dental College and Hospital, Chennai-3 having residence at NO. 24 GA, Tharangini apartment, Arunachalam road, Saligramam, Chennai – 93 (herein after referred to as the 'Researcher and Principal co- investigator')

And

Dr. V.P KIRUTHIGA aged 32 years currently studying as Post Graduate student in the Department of Prosthodontics and Crown & Bridge, Tamil Nadu Government Dental College and Hospital, Chennai-3 having residence address at No.16/7 Vallalar Nagar 1st Street, Cuddalore-1. (herein after referred to as the 'PG/Research student and Co- investigator').

Whereas the 'PG/Research student as part of her curriculum undertakes to research on the study titled **"EVALUATION OF VON MISSES STRESSES IN FOUR BONE TYPES IN AN IMPLANT SUPPORTED MANDIBULAR POSTERIOR CROWN BY 3D FINITE ELEMENT ANALYSIS"** for which purpose the Researcher and Principal investigator shall act as Principal investigator and the College shall provide the requisite infrastructure based on availability and also provide facility to the PG/Research student as to the extent possible as a Co-investigator.

Whereas the parties, by this agreement have mutually agreed to the various issues including in particular the copyright and confidentiality issues that arise in this regard. Now this agreement witness as follows:

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9. If any dispute arises as to the matters related or connected to this agreement herein, it shall be referred to arbitration in accordance with the provisions of the Arbitration and Conciliation Act, 1996.

In witness whereof the parties herein above mentioned have on this the day month and year herein above mentioned set their hands to this agreement in the presence of the following two witnesses.

College represented by its Principal	Guide	Co-guide

Witnesses	PG Student

1.

2.

MY ACKNOWLEDGEMENT AND SINCERE THANKS TO MY GUIDE

With immense pleasure and honour I take this opportunity to express my humble and heartfelt gratitude to my mentor, a relentless source of inspiration and dissertation guide **Dr.C.SABARIGIRINATHAN,M.D.S.,** Professor and head of department, Department of Prosthodontics, Tamil Nadu Government Dental College and Hospital, for his able guidance and support. I am grateful for his help at various stages of the dissertation. Without his help this dissertation would not have come out in a befitting manner. Each word said to describe the experience as his student, which was a boon in disguise, would be an understatement. His esteemed and able guidance made this dissertation a possibility. His dedication to work which made us realize the worth of discovering our own capabilities. His unprecedented calm and patient personality, an unfailing, caring and understanding demeanour made each endeavour easier.

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I am extremely thankful to my chief **DR.C.SABARIGIRINATHAN,M.D.S.,** Professor and Head of the Department and my Guide, Department of prosthodontics, Tamil Nadu Government Dental College and Hospital, Chennai, I consider it my utmost privilege and honor to express my most sincere and heartfelt gratitude to my esteemed for his wholehearted support, constant guidance, help, encouragement, valuable suggestions and support he has rendered at various stages of the dissertation. I also thank him for the valuable guidance; he has given throughout my post graduation. Without his immense help this dissertation would not have come out in a befitting manner. I am thankful to **Dr.P.RUPKUMAR, M.D.S.,** Associate Professor, Co-Guide for guiding and helping me at different stages of this study.

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Above all I thank the **LORD GOD ALMIGHTY** for giving me the strength and courage to complete this monumental task.

CONTENTS

LIST OF ABBREVIATIONS

Sl.No	ABBREVIATION	EXPANSION
1	FEA	Finite element analysis
2	CATIA	Computer aided 3 dimensional interactive application
3	ABAQUS/CAE	A backronym with an obvious root in computer aided engineering
4	.IGES	Initial graphics exchange specification
5	.INP	Input
6	.ODB	Object database
7	.STL	Stereolithography
8	3 D	3 Dimensional
9	CAD	Computer aided design
10	DOF	Degrees of freedom
11	MPa	Megapascals
12	Mm	Millimeters
13	ANOVA	Analysis of variance
14	SD	Standard Deviation
15	SE	Standard Error
16	SPSS	Statistical Package for Social science
17	N	Newton

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ABSTRACT

Introduction: It has been half a decade since the invention of osseointegration by Branemark in 1952, implants seem to have taken the limelight as they provide ideal aesthetics and masticatory efficiency. However the success rate of an implant restoration depends on wide array of factors among which biomechanics play a crucial role. Taking that into account the present study was conducted to evaluate the stresses generated in and around the implant structure in different bone qualities varying in their densities.

Key words: FEA (finite element analysis), Von mises stresses, Modulus of elasticity, Loading angle, Bone density (D1, D2, D3, D4)

Aim of the study: To analyse the stress distribution patterns in the implant and surrounding bone in D1, D2, D3, D4 types of bone subjected to different loading conditions in terms of angulation of forces applied.

Materials and Methods: Four CAD models namely D1, D2, D3, D4 were fabricated using simulation software and were meshed into a finite model with nodes and elements and was finally transferred to the analysing software (ABAQUS) and a series of 16 test runs in each individual load angulation was conducted and the results of stress concentration on seven regions of interest in the implant and the surrounding structures were obtained and the data was checked for statistical significance using SPSS software

Results: The maximum stresses among the seven regions of interest Abutment, Connecting screw, Implant, Cortical bone, Cancellous bone, Co - Cr crown, feldspathic porcelain were recorded at the neck of the implant for all the

4 bone qualities irrespective of the loading conditions and the cancellous bone was recorded with the least stresses in all the four bone qualities. The stresses recorded in all the seven region of the model showed a gradual increase in stresses as the loading angulation changed from 0 degrees to 15 degrees for all the four models and there was a significant percentage difference in the increase in von mises stresses as each five degree increase in angulation

Conclusion: Within the limits of the study it can be concluded that most of the occlusal force gets concentrated as stress in the neck of the implant region and among the seven region of interest taken for account in the study the D1 bone quality shows lesser stress recordings compared to D4 bone qualities. And all the components in the four models showed a steady increase of stress when loaded at 0° and 15°. And the 0 degree load. Where the load is applied parallel to the long axis of the implant shows less stresses and a uniform distribution.

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Sources included in the report:

Comparitive evaluation of implant designs - Influence of diameter, length and taper on stress and strain in mandibular segment - A Three dimensional finite element analysis.pdf (D34287779)

Instances where selected sources appear:

5

CERTIFICATE – II

This is to certify that this dissertation work titled "**EVALUATION OF VON MISES STRESSES IN FOUR BONE QUALITIES IN AN IMPLANT SUPPORTED MANDIBULAR CROWN BY USING 3D FINITE ELEMENT ANALYSIS.**

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ABSTRACT

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INTRODUCTION

"No one should die with their teeth sitting in a glass of water"

- Dr Per Ingvar Branemark

True to the words of Dr Branemark, the number of implants being utilised by dentulous and edentulous patients has been in a steady state of increase in countries all over the world and a 12- 15% increase is expected in the future[1]. Implant dentistry has evolved into an evidence based clinical science with documented research to validate previously unsupported clinical practise procedures. The tremendous expansion of knowledge in this field has created new ideas and terminology that is redefined based on new principles.

The goal of modern dentistry is to restore the patient to normal contour, function, comfort, aesthetics, speech and health. And the uniqueness of implant dentistry is in its ability to achieve this goal irrespective of the atrophy, disease or injury to the stomatognathic system[2].

Of recent years, biomedical engineering has gained much attention in dental implantology, particularly in terms of design optimisation and also has widened the dentists view on diagnosis, treatment planning, and rehabilitation in patient care. Biomechanics as a discipline deals with the analysis of biologic structures to various mechanical conditions to which it is subjected in a living environment as well as the various prosthetic restorations which coexist with them by utilising engineering principles[3].

Finite element analysis (FEA) is an ingenious tool used to evaluate biomechanical characteristics of different types of dental implants. The literature reflects that it has been widely used to model the design and functionality of dental implants and predict features of design optimization[4]. Finite element analysis was initially developed in the early 1960’s to solve structural problems in the aerospace industry, Which was further utilised to solve problems in heat transfer, fluid flow, mass transport and electromagnetics. Later it was adopted by the field of dentistry, proving its efficiency to be better than photoelastic study in terms of easy modelling and more defined stress analysis[5].

A key factor for success or failure of a dental implant is the manner in which stresses are transferred to the surrounding bone[6]. Though forces on teeth are applied as occlusal bite force during chewing cycles, the surrounding structures such as the tongue and perioral musculature also exert a slight but continuous transverse force on the teeth. Vertical and oblique loads from mastication bring about axial forces and bending moments and results in stress concentration in the bone as well as in the implants[5]. Forces and moments transferred from implants to the surrounding bone depend on angle of loading, the interfacial zone, implant geometry, the prosthesis type, and the quantity and quality of the surrounding bone[7]. Researches can calculate stress distribution and the related strain rates and displacements in the surrounding bone and the implant body by FEA.

Bone metabolism is regulated by two mechanisms namely, hormonal and biomechanical. And research implies that among these the biomechanical regulation plays a major role as it can maintain bone mass even in situations of calcium demand[8].The osteocytes are extremely sensitive to mechanical forces and through series of steps initiates bone formation when the stresses are kept at a ideal[9,10]. However when the stresses are at pathological levels, the bone cells stimulate the cytokines and cause bone resorption. So when the bone is stressed above the physiologic overload zone it may actually lead to the failure of implant and cause marginal bone resorption[11].

The external (cortical) and internal structure (trabecular) of the bone may be described in terms of quality or density which mirrors a number of biomechanical properties, such as strength, modulus of elasticity, osseointegration and stress distribution surrounding a loaded dental implant[12,13]. The strength of bone is directly proportional to its density. As the density increases the load bearing capacity of the bone also increases and an actual 10 fold increase in strength may be depicted as one moves from D1 to D4 bone types which naturally improves the success rate for dense bone qualities and the literature also reveals that the implants placed in anterior mandible are more likely to be successful than those placed in maxilla of poor bone qualities[14].

A tooth is designed for vertical loading, likewise the implants are also designed to withstand forces which act parallel to its long axis. An axial load imparts a compressive stress without any lateral force component and hence they can be borne well than the shear component of force which results due to a lateral load, since the bone is strongest to compression, 30% weaker to tensile stresses and 65% weaker to shear stresses[15]. So offset loads of any form, be it a vertical cantilever, force acting on an angulated abutment or due to occlusal prematurity may generate shear forces and may weaken the integrity of the restoration.

Hence in this study, the finite element method is used to evaluate stresses in an implant supported mandibular crown along with its surrounding structures in four different bone types subjected to four different offset loadings.

AIMS:

To evaluate and compare the amount of stresses in implant supported crown and the surrounding structures in four different bone qualities in different angles of loading.

OBJECTIVES:

1. To evaluate the amount of stress transmitted to the supporting structures by loading an implant supported crown axially and different angulations

   a.5°

   b.10°

   c.15°

   d.90°

2. To evaluate the amount of stress distribution in four different bone qualities

   a.D1

   b.D2

   c.D3

   d.D4

3. To evaluate the amount of stress within the implant components,

   a. Abutment
   b. Connecting screw
   c. Implant body
   d. Cobalt chromium crown
   e. Feldspathic porcelain
   f. Cortical bone
   g. Cancellous bone

REVIEW OF LITERATURE

A.M Weinstien et al (1976)[16]

He employed a two dimensional plane stress finite element analysis on porous rooted dental implants. The outcomes of this analysis were correlated with values from mechanical tests executed on actual implant specimens. The study was concluded stating that the model provided uniform distribution of stresses around the implant and a positive correlation with the mechanical tests.

A.M Weinstein, J.J Klawitter, S.D Cook (1980)[17]

In their study, 3D Finite element analysis was used to determine the implant bone interface characteristics of bioglass dental implants, which was further confirmed by comparison with the results of mechanical testing carried out on animal specimens. Utilising the Finite element model with soft tissue interface properties, the predicted load - displacement characteristics were compared and was in good agreement with measured value reported for type II bone implant interface. Interface elastic moduli were determined for the bonding layer conditions and were shown to be declining as the percentage of tissue attachment improved**.**

S.D Cook et al (1981)[18]

A three-dimensional finite element analysis (FEA) was used to analyse the effect of young's modulus on stresses in tissues surrounding implants comprising of two different materials. The use of implants coupled with a natural tooth resulted in a decrease of stresses in tissues around the teeth whereas removal of the bridge resulted in a considerable increase in stress levels around and within the implants studied and finally, the magnitude and distribution of stress in tissue around blade type LTI carbon implants were found to be near normal physiological stress levels than those around aluminium oxide implants.

Cook et al (1982)[19]

They studied 3D FEA models of 3 cylindrical porous rooted implants in order to evaluate their biomechanical response, which were verified with experimentally determined values for the same three implants retrieved from dogs mandible after two years of function and histologic analysis of the same indicated that the assumption of a direct bone - implant interface was not a better representation for these kind of porous rooted endosteal implants.

Borchers L. Reichert P (1983)[20]

He evaluated the distribution of stresses in the bone around a ceramic dental implant using a 3D finite element model, which was subjected to axial and non- axial loading at different phases of development of the interfacial zone. And the study was concluded that the crestal regions recorded the highest stress concentrations when loaded in a transverse direction and in spongy bone.

Ragnar Adell, Bo Eriksson, ULF Lekholm, Per Ingvar Branemark, Torsten Jemt (1990) – <u>A 15 year follow up study</u>[21]

This study was conducted on 700 patients both male and female with a age group that ranged from 19 to 79 years in whom 4,636 implants were placed and reviewed through annual clinical and radiographic evaluation over a period of 15 years and the prosthesis stability was 95% for maxilla at 5 and 10 years whereas for the mandible it was a solid 99% throughout the study period and they also obeyed the specific criteria given by Albrektson et al.

Reiger et al (1990)[22]

He conducted studies on six endosteal implants with similar modulus of elasticity which were loaded axially in order to compare the stresses among them using two dimensional finite element analysis method. And based on the results, a cylindrical implant design directed most of the applied axial load to the apical bone while the tapered design provided better stress distribution**.**

Bertil friberg, Torsten jemt, ULF lekholm[23]

A three year period retrospective study

A total of 899 patients were involved in the study with 4641 implants placed in both upper and lower jaws with the study group comprising a mean age group of 57.5 years and the patients were regularly followed up from the first surgical procedure to the final restoration. And the failure rates accounted to about 1.5% of which most of the failures were in the maxillary jaw with poor bone qualities. And length seemed to play a role too, as many failures occurred in short implants.

Meijer H J et al (1992)[24]

He, along with his colleagues conducted a two dimensional finite element analysis on two implants connected by a bar on a mandibular model in order to evaluate the stresses within and surrounding the design, taking into consideration the height and length of the mandible along with various superstructures. And it was concluded that the length of the mandible had no influence over the stresses, whereas the height of the mandible largely influenced the stress patterns. And the stresses were highly concentrated near the neck of the implant and in shorter length implants and also a model with a solitary abutment showed a more uniform stress distribution when compared with a model with connected abutments**.**

Rho et al (1993)[25]

This particular study utilised ultrasonic and micro tensile testing techniques to determine the Young's modulus of dried histologic samples of trabecular and cortical bone cut to same sizes. The results obtained from both testing techniques shows that the mean trabecular Young's modulus was significantly less than that of cortical bone. However the fact that the specimens were actually dried before testing should be considered and so a higher young's modulus values should be expected in invivo results.

Meijer H J et al (1993)[26]

They conducted a 3 Dimensional FEA on two implants placed in between the interforaminal regions which were either solitary or connected in order to study their stress distribution patterns. The computer aided design was built from data obtained from 3D slices of a cone beam computed tomographic representation of a single human mandible, which was subjected to different offset loads. The extreme principal stresses were always located around the neck of the implant and when loaded with oblique bite forces. While vertical bite forces were recorded with the least stresses.

Meijer H J et al (1996)[27]

They calculated the stress distribution around dental implants in an edentulous mandible by means of a three dimensional finite element model of the anterior part of the jaw. They determined the most extreme stresses in the bone were always located around the neck of the implant. In the case of uniform distribution of the loading there were more or less equal extreme principal stresses around the central and lateral implants. They also found that if the load was not uniformly distributed on the super structure then the implant that was nearest to the place of loading showed the highest stress concentration.

Mericske – stern R et al (1996)[28]

This is an in vivo study conducted in implant supported overdentures using piezoelectric transducers that can register force in all 3 dimensions. This study was actually conducted to compare between a rigid bar, round bar and telescopic design supporting an overdenture. It was seen that force magnitudes of chewing and grinding were identical for all three anchorage devices. However differences were observed amongst bars and telescopes with slight increase of stresses for the telescopic design.

Papavasiliou G et al (1996)[7]

Used three dimensional finite element analysis model to scrutinize various parameters in regards to bone implant interface. Stress circulation patterns were compared and interfacial stresses were scrutinised precisely at four heights along the bone implant interface. And was inferred that oblique loads amplified stresses 15 times. Conditions for bone microfracturing were associated with oblique loads, high occlusal stresses and the absence of cortical bone.

K. Snauwaert et al, J Duyck, D van Steenberghe, M Quirynen, I Naert[29] (1999)

<u>A 15 year follow up cohort study[29]</u>

This study was conducted in 1315 patients both males and females who were in various stages of partial and full edentulousness. They were further subdivided into the ones with healthy grafted bone and patient who underwent irradiation therapy for head and neck cancers. Among the total of 4971 Branemark implants there was failure rate of 19% in the compromised individuals and a failure rate of 5.9% in the non-compromised group of the total implants placed and most of the failures were recorded in the maxilla and the implants of shorter lengths.

Roxanna Stegariou et al 1998[30]

They analysed stress distribution patterns in a fixed implant prosthesis design with two implants abutments and the bone surrounding it with different materials for crown superstructures with axial and non-axial loading of forces. And a total of 3 different restorative materials were involved in the study and was concluded that the maximum stresses were generated from non-axial loading. Further when the materials were compared more stresses were recorded in acrylic followed by composite resin, porcelain and gold. And they also state that since acrylic takes much stresses the implant gets loaded less and hence acrylic as a restorative material during the initial restorative phases may prove to be good.

Esposito M et al 1998 [31]

<u>Meta-analysis</u>:

This is a literature review which assess 73 articles in order to acquire knowledge about osseointegration, failed and failing implants, their success rates, and periimplantitis and concluded that both early and late failures were predominant in maxilla than mandible with a 3 time increased failure rates than the lower jaw and both surgical trauma and anatomic conditions accounts to the total implant losses of 3.6%. And the failures due to periimplantitis often occur in combination with other factors such as bone quality, quantity and a number of clinical variations.

Roynesdal AK et al (1998)[32]

Observed the clinical outcome and marginal bone resorption of three different endosseous implants positioned in the anterior mandibular region. After a three year period it was established that titanium plasma sprayed symmetrical implant had a less favourable scenario than non-coated implants used in the study.

Teixeira ER et al (1998)[33]

They conducted finite element analysis in implants with more concentration on the distribution of stresses in the periimplant region. The researchers innovated the 3 D model by unification of elements which were placed farther, thereby reducing the number of elements, time duration and the computer memory. The results propose that it is possible to develop a replica of FEA implant model of the mandible with less range and fewer elements without actually altering stress distribution.

David L.Cochran (1999)[34]

They performed an investigation on endosteal implants which varied in their surface modifications. The study reveals that implants with rough surfaces proved to be more successful than the ones with smooth surface topography. But in single implant crowns there was no significant difference in success rates between the two types. And also the implants placed in mandible had better success rates than those in maxilla.

Vollmer et al[35] (2000)

This study was conducted to compare the finite element analysis with that of in vitro loading conditions. The FEA part of the study was conducted by constructing a mandibular model from the results obtained from a CBCT in order to analyse stress, strain and deformation. And the results were concluded stating that there was a good correlation between the two methodologies. So the FEA method can be used to analyse biomechanics since it is both accurate and a cost effective method.

Beat r. merz et al (2000)[36]

This research presents a comparison between morse taper and the butt joint connections between an implant and an abutment. They conducted a non-linear dynamic 3D FE analysis in order to compare them with long term fitness in mind. Hence they concluded that the conical abutment condition seemed to be better in comparison with the other type of connection.

J P Geng et al (2001)[37]

This study is a review of literature about finite element analysis and its applications in dentistry. The reviewers explains in detail about geometric modelling of the living structures and dental implants, the various loading and boundary conditions. He has tabulated the various material properties and their significance and the qualities of bone such as its non homogenicity. And finally the advantages and limitations of the method.

Baris Simsek et al (2004)[38]

Conducted a study to evaluate the effects of different inter implant distances in different implant systems on stress distribution in the bone around titanium implants under vertical, horizontal and oblique loads in buccolingual, mesiodistal, and vertical directions in the posterior mandibular region by finite element analysis and reported that 1 cm inter implant distance was the optimum for two implant fixtures adjacent to each other, they found out that 2 cm distance increased the tensile stresses under vertical and mesiodistal loads whereas in the 0.5 cm distance there was an increase in compressive stresses in all three directional forces.

J P Geng et al[39] (2004)

This study compared force distribution between stepped screw and cylindrical screw implants using 2 dimensional finite element analysis with axial, transverse and oblique loading conditions with varying elastic modulus for cortical bone and trabecular bone and the maximum von mises stresses generated were found to be 17.9% lower in stepped screw implant area in the trabecular bone which is because of the low stiffness of the stepped screw implant.

Gary R O' Brien et al, Aron Gonshor, Alan Balfour (2004)[40]

A 6 year prospective study

He conducted studies on a total of 620 dental implants. Out of the total number of implants, 386 implants were a prototype HA coated implants which he claims to be a unique stress diversion surface system developed by a combination of mathematical modelling, digital radiography (isodensity) and FEA and the remaining 234 were grit blasted commercially available implant systems which were placed in both augmented and non- augmented bone types and the prototype obtained a slightly higher survival rate (96.6%) than the existing implant system and the maximum stress concentration were at the crest of the ridge.

Sevimay et al, (2005)[41]

In this study, a 3 D Finite element analysis was conducted on an implant crown embedded in the mandible simulating four different bone qualities. Each model was subjected to axial loading in order to evaluate the stress distribution patterns. The neck of the implant had the highest stresses and the D1 bone quality recorded the least stresses followed by a gradual increase of stresses as one moves towards D4.

Porter J et al, (2005)[42]

Meta analysis:

This is a review of literature about success and failures of dental implants. The reviewers conducted an extensive study on articles which mainly centred around meta analysis and multicentre studies and they gleaned that the main factors for implant survival was based on patient factors such as age, bone quality and quantity, oral hygiene, systemic disease, smoking, parafunctional habits, the biomechanical factors such as prosthesis design, length, position and the direction of loading of implants.

Omar lutfi koca et al (2005)[43]

They conducted a study utilising a finite element model of posterior mandible with varying crestal bone heights (4, 5, 7, 10 and 13 mm) with a vertical loading protocol, and stated that in supporting bone levels of 4 and 5 mm, the maximum stresses were localised in wide areas and particularly in the neck of the implant whereas in 7, 10, 13 mm stresses were localised in narrow areas and the maximum von mises stresses were recorded on the palatal cortical bone for all irrespective of the crestal bone levels.

M. Karl et al (2007)[44]

They analysed six different implants supported FPD's with strain gauges on the pontics of the prosthesis on vertical loading and concluded that the same magnitude for both three unit and five unit prosthesis were derived through FEA results. However the in vivo measurements revealed a higher stress values for the five unit prosthesis. The strain gauge values show low von misses stresses for the cement retained prosthesis than the screw retained prosthesis and an overall reduction in von mises stresses in the apical area of the implant for all the groups.

Heoung – jae chun et al (2006)[45]

This study investigated the effect of different abutment types on stress distribution in bone with inclined loads using finite element analysis. Separate models with internal hex and external hex were modelled to study the effect of abutment types on stress distribution. It was found that internal hex implant system generated lowest von mises stresses for all loading condition because of reduction of bending effect by sliding in the tapered joints between implant and abutment concluding the abutment type had significant effect on stress distribution in bone.

Jose henrieque rubo et al, (2008)[46]

In this study finite element analysis was conducted on an implant supported prosthesis consisting of five implants with a bar framework loaded in a vertical direction. And it was concluded that each increment of 5 mm cantilever length augmented the stresses by approximately 30- 37 % on the compact bone. They also observed that stiffer the cancellous bone the more stress they seemed to bear and the cortical bone was comparatively loaded less and diminished stress was observed with increase in the length of implant.

Arun kumar et al, (2012)[47]

In this study, two implants one with a straight abutment and the other with an angulated abutment which were assigned to be in groups containing different bone qualities were assessed by loading with a static load. The highest stresses were in the angulated abutment in the poorer bone qualities such as the D4 type and the cervical region was the one with greater stress Concentration compared to the rest of the design.

Marcello bighetti toniollo et al (2012)[48]

This study was conducted in two dental implants, with morse taper connection with same diameter and varying lengths of 11 and 13mm lengths. A FEA was conducted to find out the effect of stresses and stability in terms of comparison between those fixtures that were implanted into the cortical bone and those implanted in trabecular bone. And the results of the study concluded that the stability was better for implants positioned in cortical bone than those placed completely in trabecular bone. Yet implants placed slightly below the neck of the crest has a added advantage of lesser crestal bone loss.

Miyuki omari et al (2015)[49]

This study was conducted in order to compare between the values obtained from FEA analysis and an experimental model which were subjected to forces by Instron load analyser. A FEA model and an actual physical model was constructed and subjected to same loading conditions and the data obtained was statistically analysed by availing coefficient of variance. The results report a variance of 5 – 10% between the two methods.

Fellippo Ramos Verri et al (2015)[50]

The researches plotted the von misses stresses, principal stresses and displacement in monocortical and bicortical implant placement in the anterior maxilla using 3D finite element method with loading at different directions and depicts bicortical technique to be best from a biomechanical perspective. Oblique loads augmented the stresses in all techniques especially in the fixation screws and the cervical area of the implant.

Julius Maminskas et al, al Girdas Puisys, Ritva Kuvoppala, Aune Raustia Gintaras Juodys Balys (2016)[51]

<u>A systematic review</u>

This study was conducted by collecting data from online search performed on MEDLINE and EMBASE databases for a past five year period from the time the study was done. The study included finite element analysis which varied in their analyses ranging from influence of length and diameter of implants, the load characteristics, the implant abutment connection, different bone densities, osseointegration, on generation of stresses, strains and other biomechanics and concluded that offset forces acting on implants, cantilever in the system, misfit in the implant abutment connection, the magnitude of force and the difference in properties of restorative materials brought about periimplant strain.

Camilla lima de Andrade et al[52] (2016)

This study conducted a three dimensional finite element analysis on a single implant design with a zirconia superstructure in four different types of implants which varied in their implant abutment connection on thread design and implant body shapes and the bone design was simulated such that of posterior maxilla with poor bone quality. The results states that the maximum stress and strains were present in the posterior maxilla models with external hexagon connection irrespective of the implant body types.

MATERIALS AND METHODS

The study was performed to determine stress distribution patterns at the mandibular first premolar region in 4 implant supported crown models in different bone qualities (D1,D2,D3,D4) which were subjected to loading in 4 angulations, 0°, 5°, 10° and 15°. And each model was subjected to four test runs in each of the four angulations by three dimensional finite element analysis.

INSTRUMENTS USED FOR THE STUDY:

1. **Personal computer configuration:**

   CPU: 4

   PROCESSOR: Intel core® Core™ i5 760 processor, 2.80 GHZ

   MEMORY CAPACITY: 8 GB Ram

   GRAPHIC CARD: NVIDIA (2GB)

2. **Software specification:**

   FOR CAD MODELLING: Catia v5 R 19

   FOR MESHING: Hypermesh v 13.0

   FOR ANALYSIS: ABAQUS 6, 14 - 2

3. **Model fabrication:**

Sl.No	Materials	Commercial name
1.	Implant	Adin Touareg™ S
2.	Vernier caliper	Mitutoya vernier caliper
3.	Poly vinyl siloxane impression material	DPI photosil soft putty
4.	Polyvinyl siloxane	Kulzer variotime bite registration paste
5.	Vaseline petroleum jelly	Unilever
6.	Extra oral scanner	Imix ceramil motion 2

INPUT FOR THE CAD MODEL:

IMPLANT MODEL:

Implant: Diameter : 4.2 mm

Length : 10 mm

See fig.

BONE MODEL:

Sl.No	**Bone quality**	**Bone model**
1.	D1	Homogenous compact bone
2.	D2	Thick compact bone(2mm) surrounding a core of dense trabecular bone
3.	D3	Thin layer of cortical bone(1mm) surrounding a core of dense Trabecular bone
4.	D4	Thin layer of cortical bone(1mm) surrounding a core of low density trabecular bone

The bone models were designed according to Lekhlom and Zarb (1985)[53] classification.

MESH MODEL:

Sl. No.	**Constituent parts**	**Numbers**
1.	Nodes	323910
2.	Elements	1691090
3.	Degree of freedom	323910×3
		1691090×3

Loading Pattern:

A total force of 300 N as a single static load is applied at the buccal cusp (150N) and distal fossa (150N) as point loads in centric occlusion at the vertical axis (perpendicular to the occlusal plane) and also as offset loads in 5°, 10°, 10° and 15°.

The models were subjected to forces from four different angulations,

1. 0 Degree angulation
2. 5 Degree angulation
3. 10 Degree angulation
4. 15 Degree angulation

BOUNDARY CONDITIONS:

Restricted in all degrees of freedom in the x, y, and z directions

MATERIAL PROPERTIES:

SL. NO	COMPONENT	MATERIAL	YOUNG'S MODULOUS (N/mm^2 /Mpa)	POISSONS RATIO	DENSITY (Ton/mm^3)
1.	Abutment	Titanium alloy	110000	0.35	4.96 e^-9
2.	Connecting Screw	Titanium alloy	110000	0.35	4.96 e^-9
3.	Implant body	Titanium alloy	110000	0.35	4.96 e^-9
4.	Crown superstructure	Co – Cr alloy	218000	0.33	9.15 e^-9
		Feldspathic porcelain	82800	0.35	2.75 e^-9
5.	Bone	Dense trabecular	1370	0.30	2.09 e^-9
		Low dense Trabecular	1100	0.30	2.09 e^-9
		Cortical	13700	0.30	2.09 e^-9

YIELD STRENGTH:

SL.NO	MATERIAL PROPERTIES	YIELD STRESS (0.2%)
1.	Titanium*	860
2.	Co – Cr alloy	500
3.	Feldspathic porcelain	380
4.	Cortical bone*	188
5.	Trabecular bone*	88 – 121

* 54

FEA:

STEPS IN FEA:

1. Pre processing

   a. Solid modelling

   b. Meshing

   c. Boundary conditions

2. Processing

3. Post processing

   a. Result interpretation

Step 1:

PRE PROCESSING:

This step includes drafting of the body to be analysed. With the details collected by measuring the actual implant and data obtained from stereolithographic image, In the present study a CAD model of the posterior mandibular segment comprising of an implant supported crown was fabricated and the periphery was plotted as x, y, z and restricted in all degrees of freedom.

Working steps:

- Constructing geometric data of the structure to be analysed
- Importing the data
- Geometry check/cleanup
- Deciding the element type
- Assigning material properties to the structure
- Loading conditions to which the model is subjected

In our study an implant supported crown (mandibular premolar crown) was designed along with the surrounding bone block (28.6mm – 16.5mm). It was designed by values obtained by two methods which were from the size and dimension of actual implant measured by vernier calliper and stereolithograhic images obtained by making putty impression of actual implant and positive models obtained with bite registration paste which was scanned with IMIX extraoral scanner. Each of the implant components (abutment, connecting screw and implant body) were drafted separately and for the crown structure the metal and feldspathic porcelain was given a thickness of 0.2mm and 1.8mm respectively and as for the shape of the premolar crown standard values of premolar tooth was assigned (wheelers oral anatomy)[55] and the bone (both cortical and cancellous) was designed according to Zarb and Lekholms classification for each of the different bone types and the boundary conditions were assigned as x, y, z for the mesial, distal, buccal and lingual, and apical sides of the model, shortly the lower part of the bone is arrested in all DOF (Tx, Ty, Tz, Rx, Ry and Rz).

Models were created using hypermesh software and later the model was imported to ABAQUS through IGES (initial graphics exchange specification) for proceeding with the analysis.

MATERIAL PROPERTIES:

FEA assigns the following mechanical properties for the material components

1. HOMOGENOUS : The properties are uniform throughout the structure.
2. ISOTROPIC : Iso – same Tropic – direction the properties remain the same in any given direction.
3. LINEARLY STATIC: The strain/deformation are proportional to the applied loads.

ELEMENT TYPE:

The models were meshed with tetrahedron elements throughout the model.

PROCESSING:

The geometric CAD data was checked for errors and a cleanup before meshing was done. The CAD model was subjected to meshing using hypermesh software. And a load of 150 N each was given at the buccal cusp tip and distal fossa (control points). So a total of 300 N[56] was applied on the premolar simulating a centric occlusion bite force and displacement seen at nodes.

WORKING STEPS:

1. Providing of a control data.

2. The different layers of the body to be analysed are represented as different areas.

3. Utilizing the hypermesh software and meshing done.

The ABAQUS Software was employed to generate input data for the stress analysis. Geometric and elastic parameters of all components were entered into the computer program.

The model thus created was given life like properties by inducing into the different layers their modulus of elasticity, poissons ratio and density. And thus the stress values were propagated when load tested.

LOADING:

The model was loaded at two points namely buccal cusp tip and distal fossa simultaneously with 150 N each as static load so as to simulate centic occlusion and every individual model was loaded from four different angles,

Which were,

1. 0 degrees (parallel to the long axis of tooth)
2. 5 degrees
3. 10 degrees
4. 15 degrees

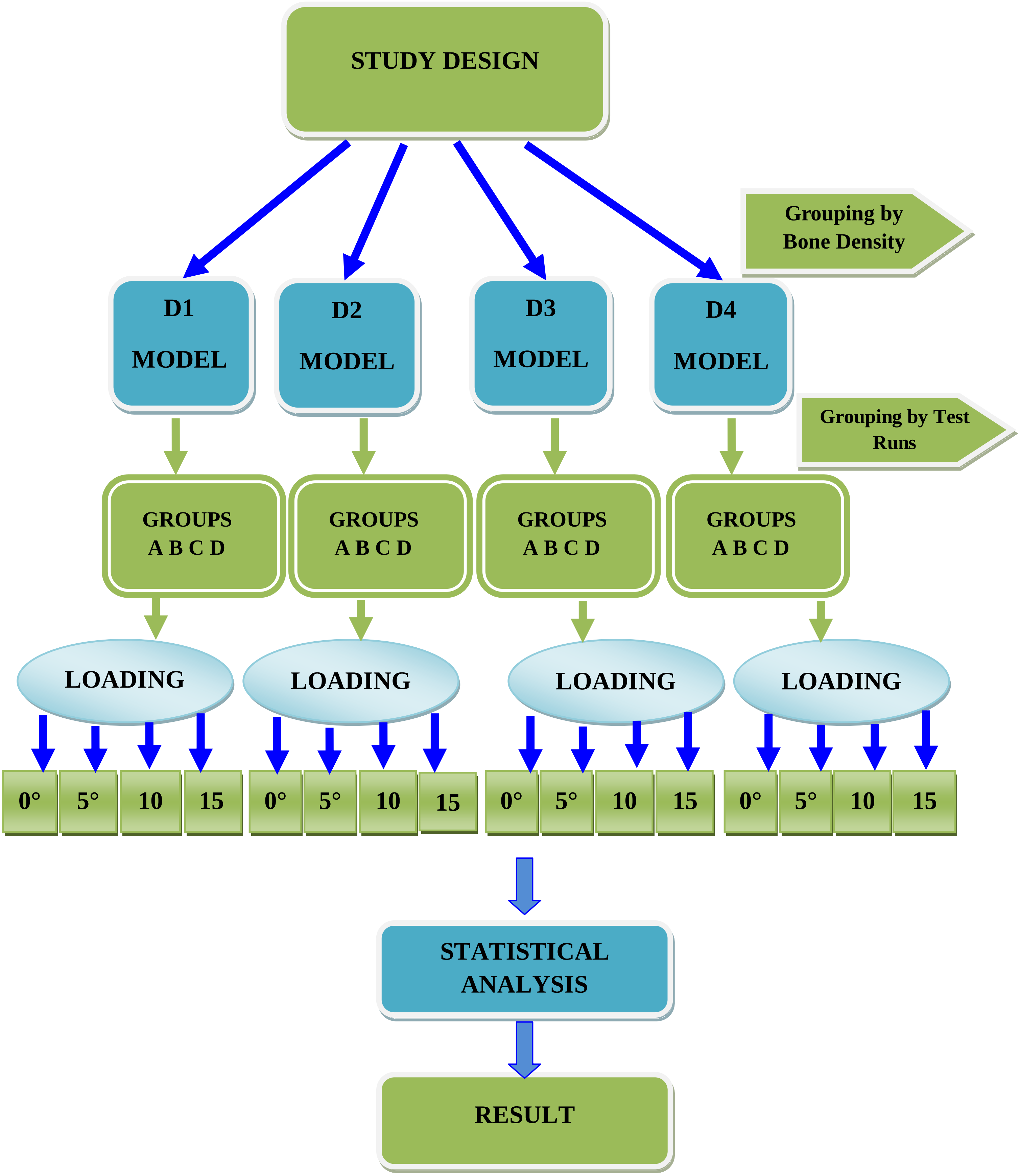
The model depicted stresses both numerically and by colour code.

POST PROCESSING:

Once the result data was obtained as numerical values and colour coded they were tabulated and analysed for computation of the results.

COLOUR CODING:

The result was given as von misses stresses in megapascals. Since Von mises are the ones which are used for ductile materials and shear stresses generated by loading at an angle and hence appropriate for the present study.


STUDY DESIGN
Grouping by Bone Density
D1 MODEL
D2 MODEL
D3 MODEL
D4 MODEL
Grouping by Test Runs
GROUPS A B C D
GROUPS A B C D
GROUPS A B C D
GROUPS A B C D
LOADING
LOADING
LOADING
LOADING
0° 5° 10 15
0° 5° 10 15
0° 5° 10 15
0° 5° 10 15
STATISTICAL ANALYSIS
RESULT

OVERVIW OF FEM:

FEA is an acknowledged methodology based on sound principles designed for engineering. Its development can actually be traced way back to Hrennikoff (1941)[57] and Courant (1943)[58] while the practical application of the method is linked with Leonard Oganesyan (1960). It is a problem solving tool which was later adapted to various fields. In simple words, FEA takes up a complex problem and breaks it into small pieces ie converts a solid structure of infinite elements into finite elements to arrive at a solution.

APPLICATIONS:

- Mechanical/aerospace/civil/automotive engineering
- Structural/stress analysis
- Static/dyanamic
- Linear/non linear
- Fluid flow
- Heat transfer
- Electromagnetic fields
- Solid mechanics
- Acoustics
- Biomechanics

TERMINOLOGY[59]:

FEM:

Any continuous object has infinite degrees of freedom. This method reduces degrees of freedom from infinite to finite with the help of discretization (ie) meshing (nodes and elements) in order to solve a complex problem.

ELEMENTS:

It is the entity which joins the nodes and forms a specific shape such as triangular, quadrilateral etc. and it is desirable that various elements can be created to offer users with the needed flexibility to meet the requirement.

NODES:

All the calculations are made at limited number of nodes. They are present in or outside the boundaries of the model to be meshed.

DEGREES OF FREEDOM:

The minimum number of parameters (motion, co-ordinates, temperature etc.) required to define position of any entity completely in space is known as DOF.

YIELD STRENGTH:

It is the stress at which a specific amount of plastic deformation is produced, which is usually taken as a 0.2% of unstressed length.

STRESS:

Stress = force/area

TYPES:

<u>Normal stress</u>: acts perpendicular to the cross section a causes compression and elongation

<u>Shear stress</u>: acts parallel to the cross section causes distortion

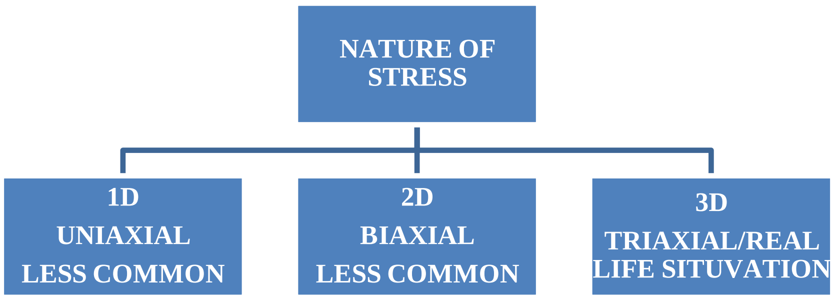
VON MISES STRESS:

It is based on distortion energy or shear strain energy (Theory of failure). These are the stresses above which the material yields or deforms. Hence recommended for ductile materials

Failure Criteria:

Shear strain energy (multiaxial loading) = shear strain energy at yield point.

NATURE OF STRESS:

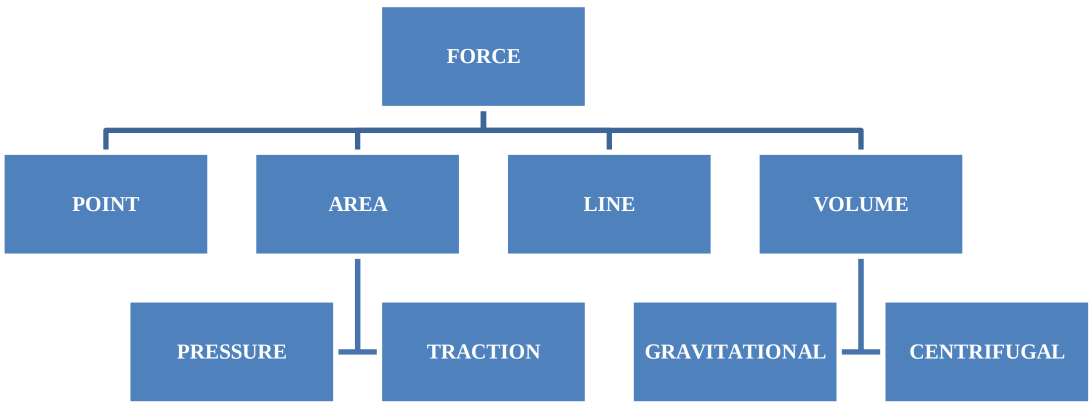


FORCE:

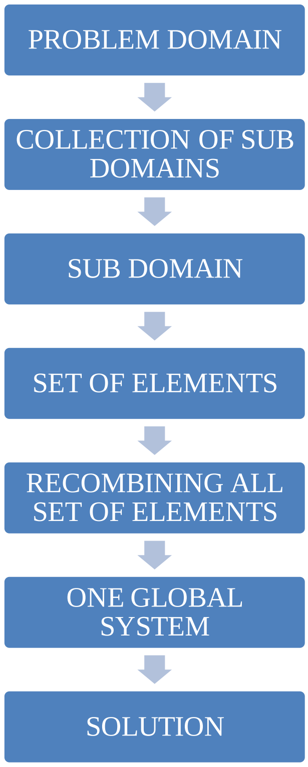
F = Mass/area

TYPES:

(Based on region of application)



CONCEPT OF FEA:



ADVANTAGES:

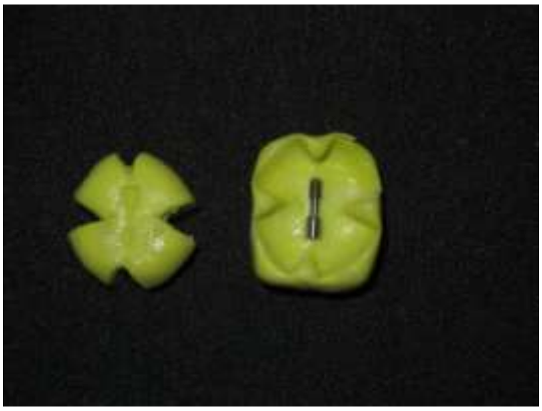
1. Increased visualisation
2. Decreased design cycle time
3. Reduction in the number of prototypes
4. Number of testing times reduced
5. Optimum design

DISADVANTAGES:

1. Precise material properties are needed for the results to be accurate
2. It entails a large amount of the computer's memory capacity and time
3. The results are generated as a large amount of numerical data, which again is a time consuming procedure to extract the required results.

FIG. 1 ARMAMENTARIUM FOR METHODOLOGY

FIG.2 IMPRSSION OF THE IMPLANT MODEL



ARMAMENTARIUM FOR STERIOLITHOGRAPHY
EXTRA ORAL SCANNER

FIG. 3. IMPLANT DIMENSIONS AND STEREOLITHOGRAPHIC MODEL

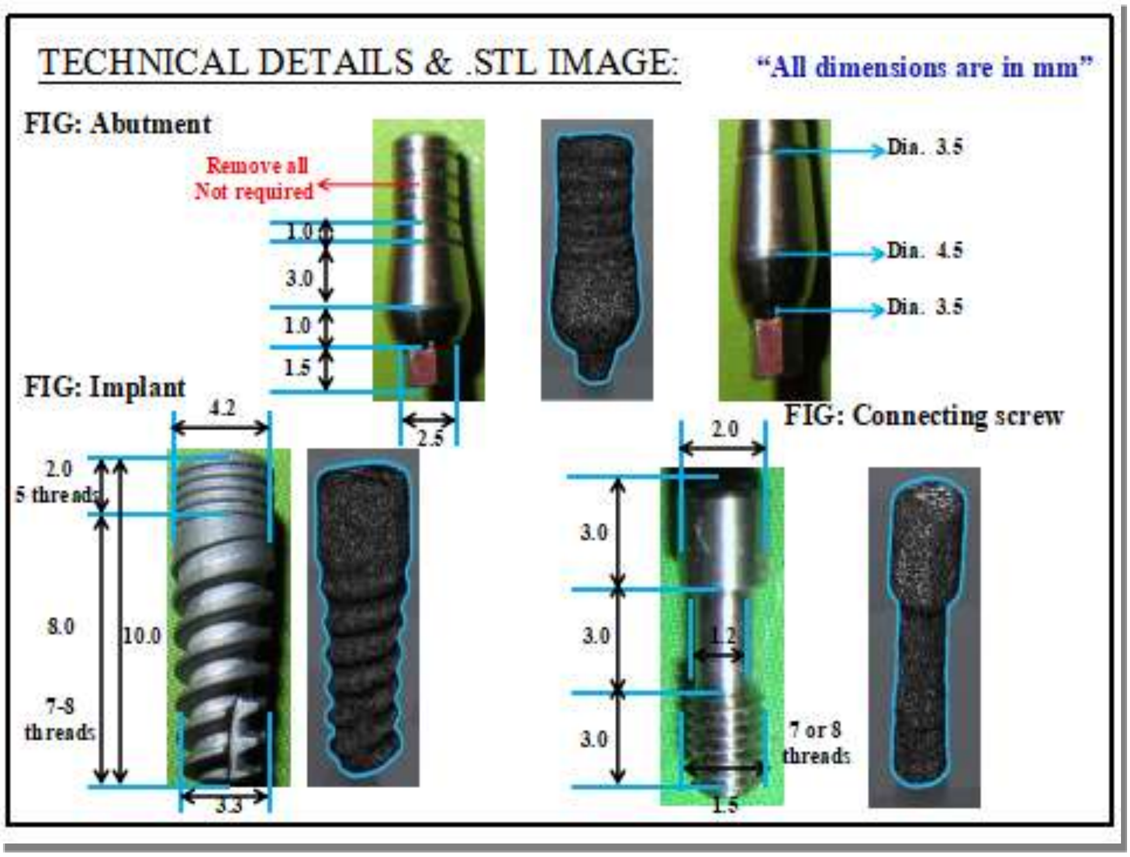
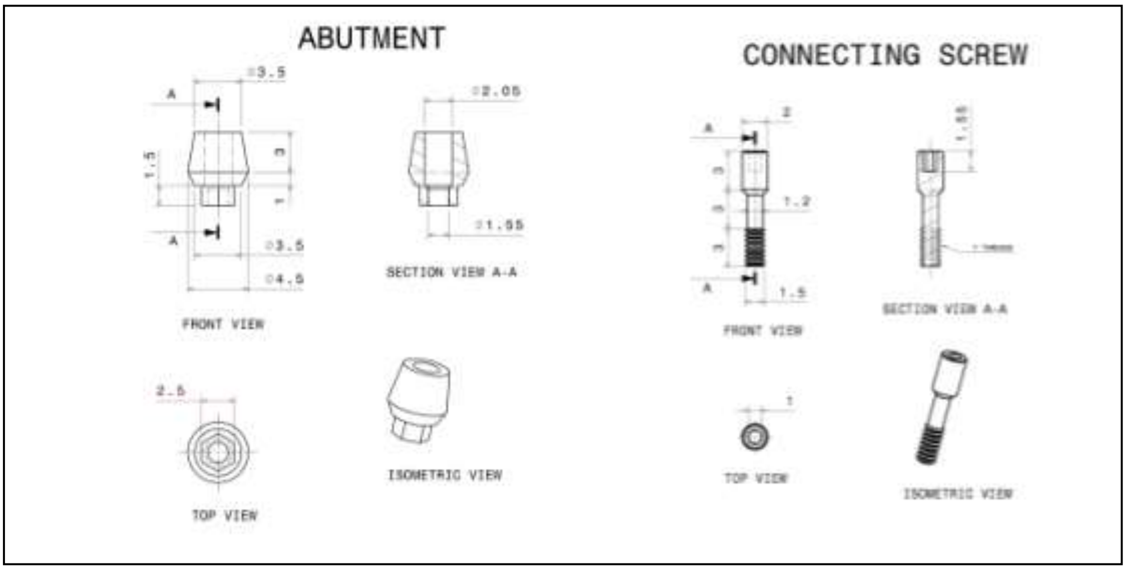


FIG. 4. GEOMETRY OF IMPLANT MODEL



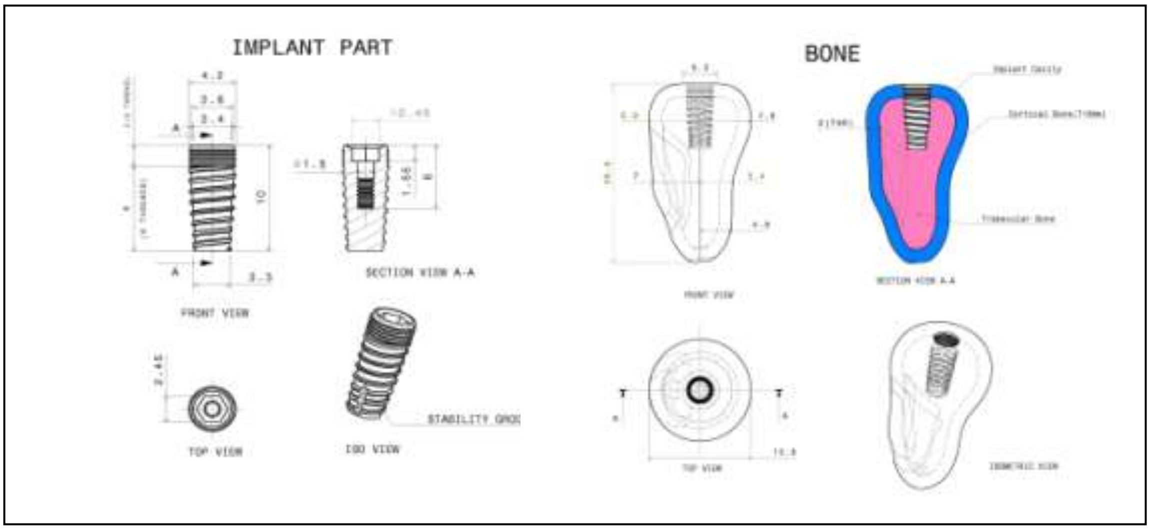


Fig. 5 CAD MODEL OF THE D1,D2, D3, D4 BONE MODELS

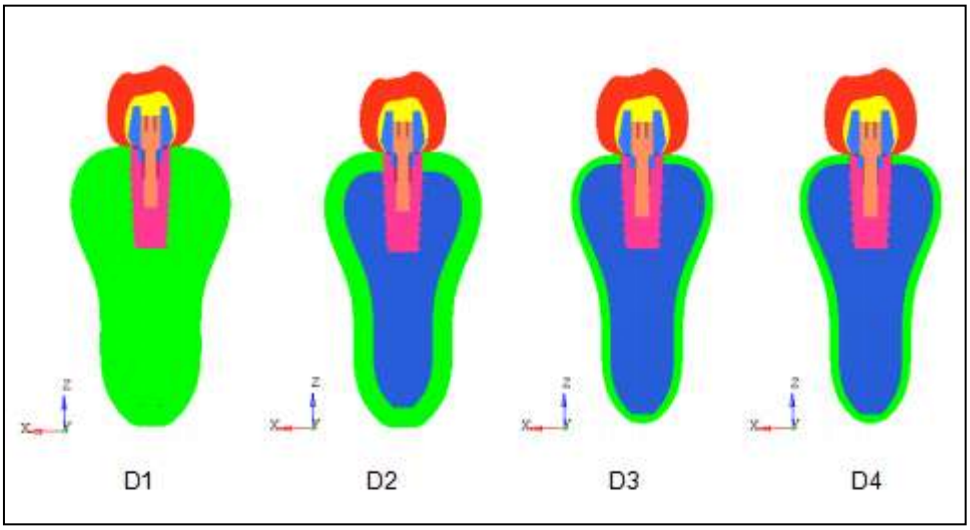
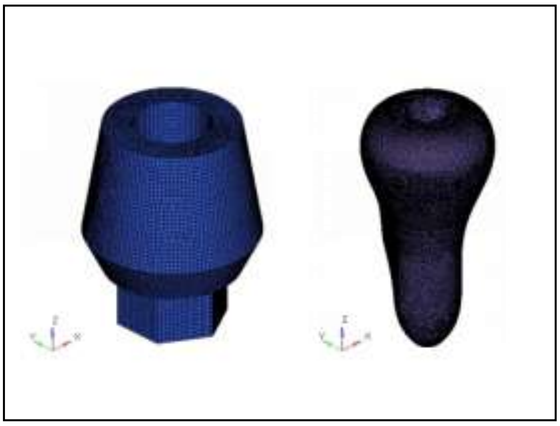
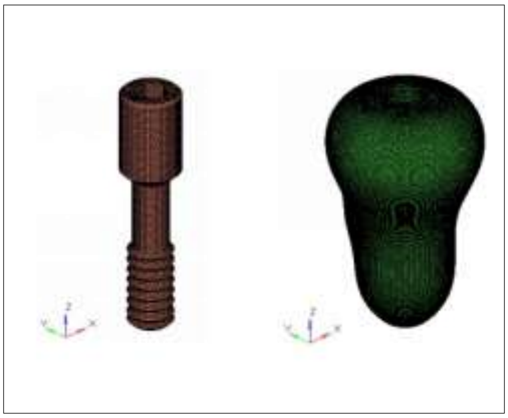
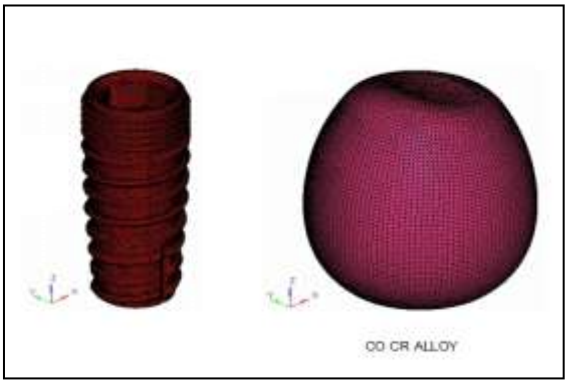


FIG. 6 MESHING OF THE BONE MODELS







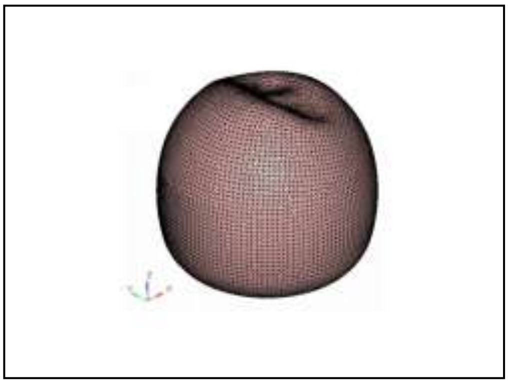


FIG. 7 LOADING OF THE FINITE ELEMENT MODEL

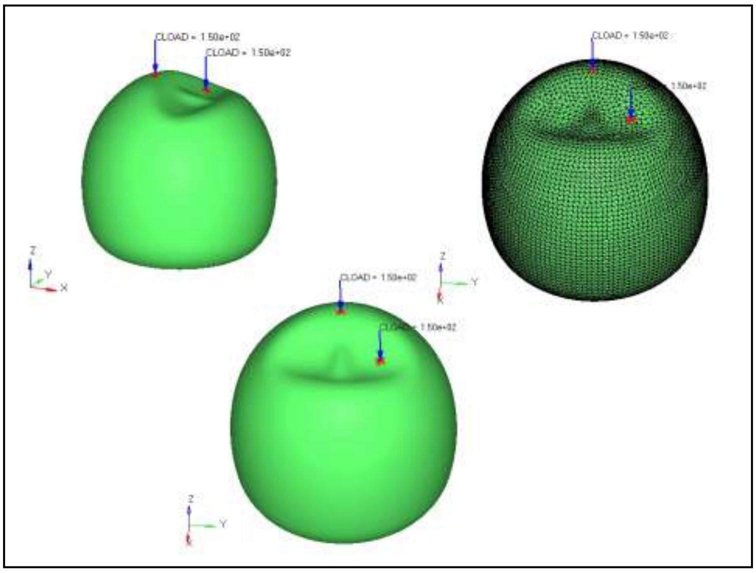


FIG. 8 : D1 LOADED AT 0 DEGREE

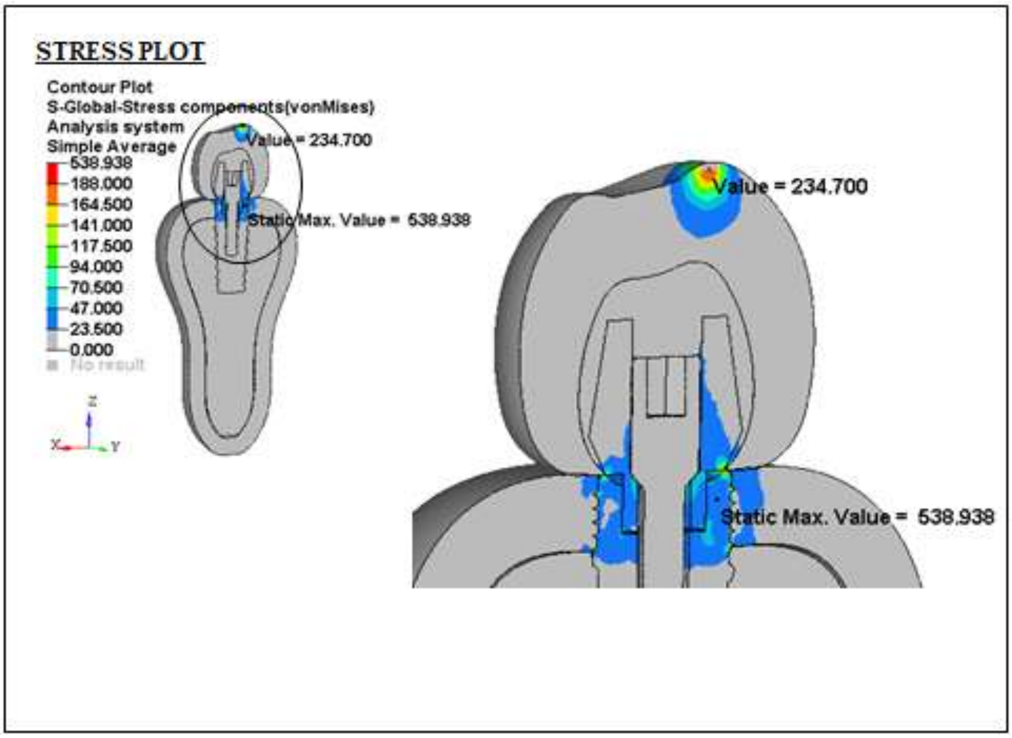


FIG. 9: D1 LOADED AT 0 DEGREE

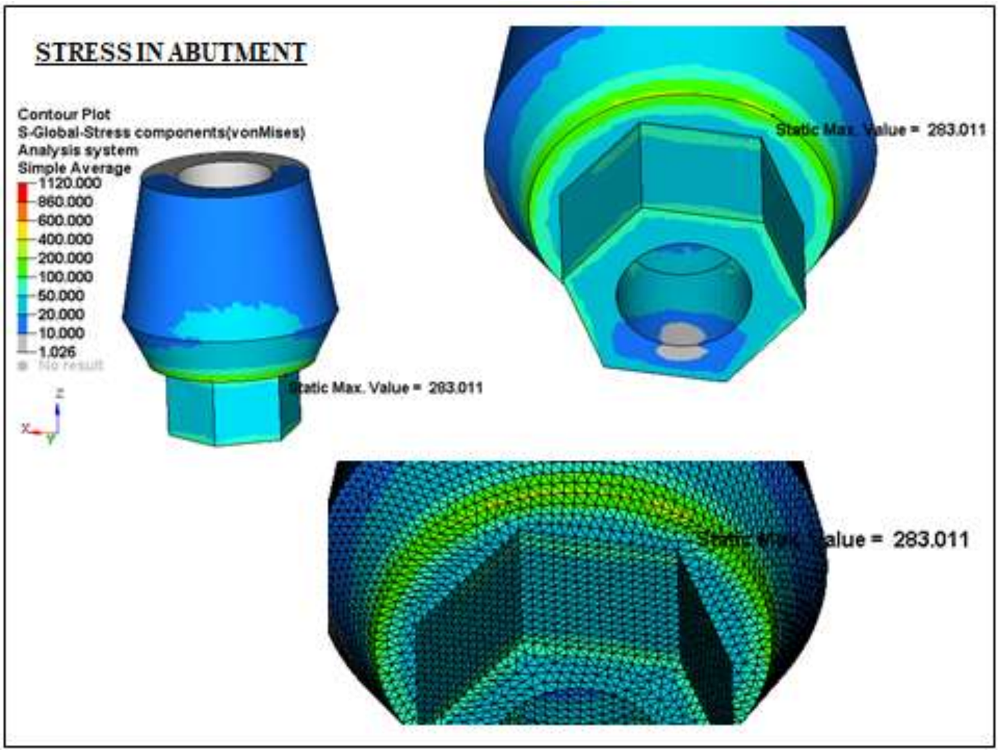


FIG. 10: D1 LOADED AT 0 DEGREE

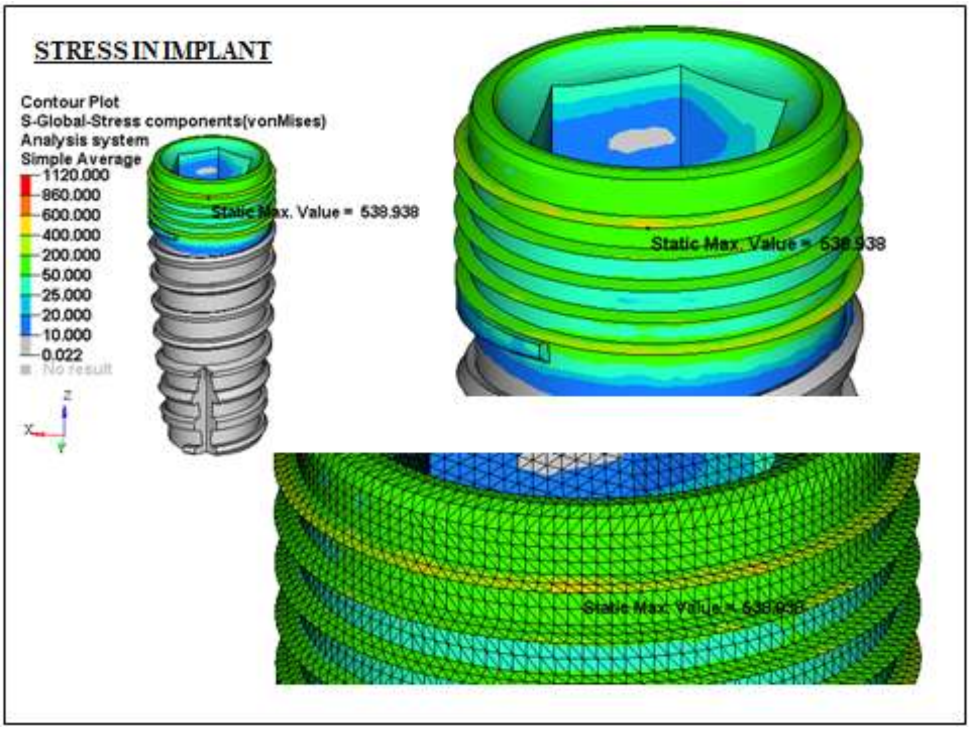


FIG. 11: D1 LOADED AT 0 DEGREE

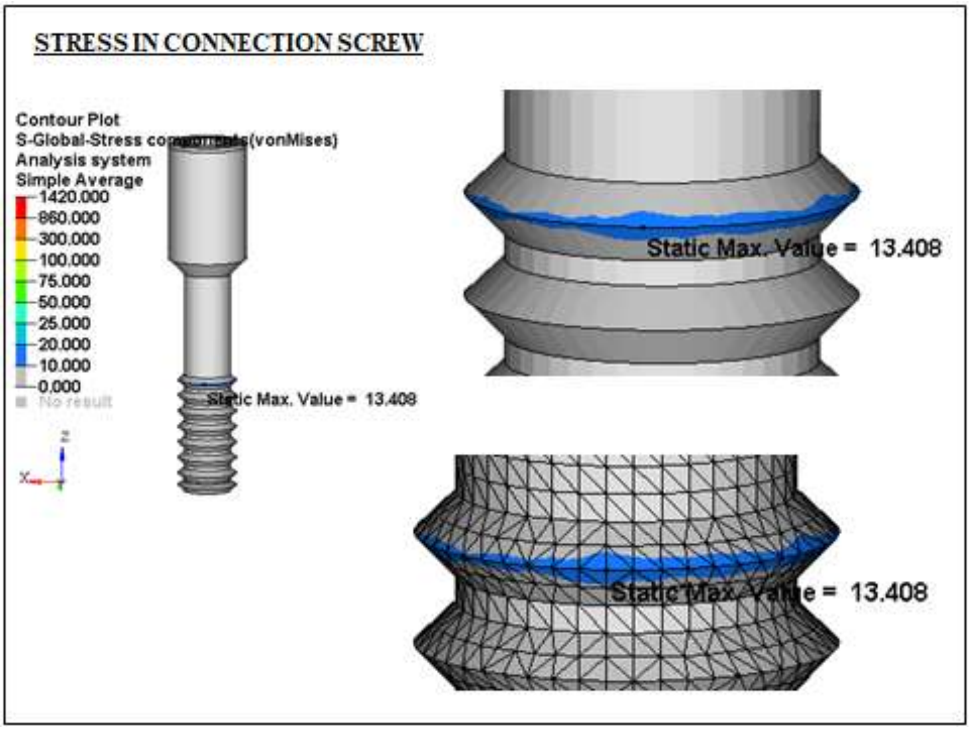


FIG. 12 D1 LOADED AT 0 DEGREE

STRESS IN CROWN CO-CR ALLOY

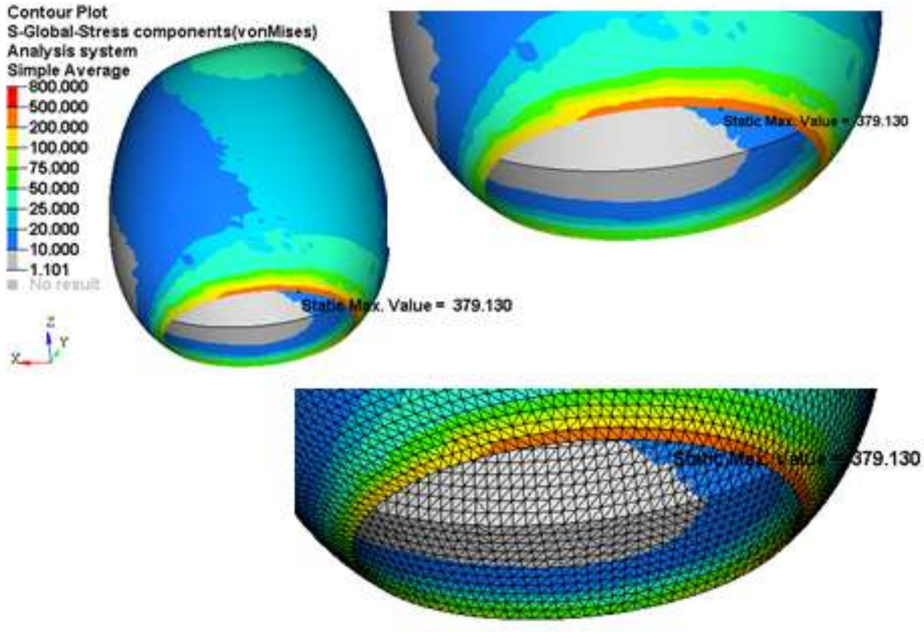


FIG. 13: D1 LOADED AT 0 DEGREE

STRESS IN PORCELAIN

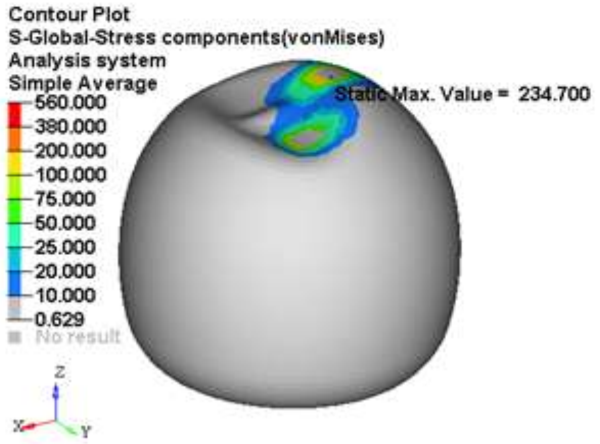


FIG. 14 : D1 LOADED AT 0 DEGREE

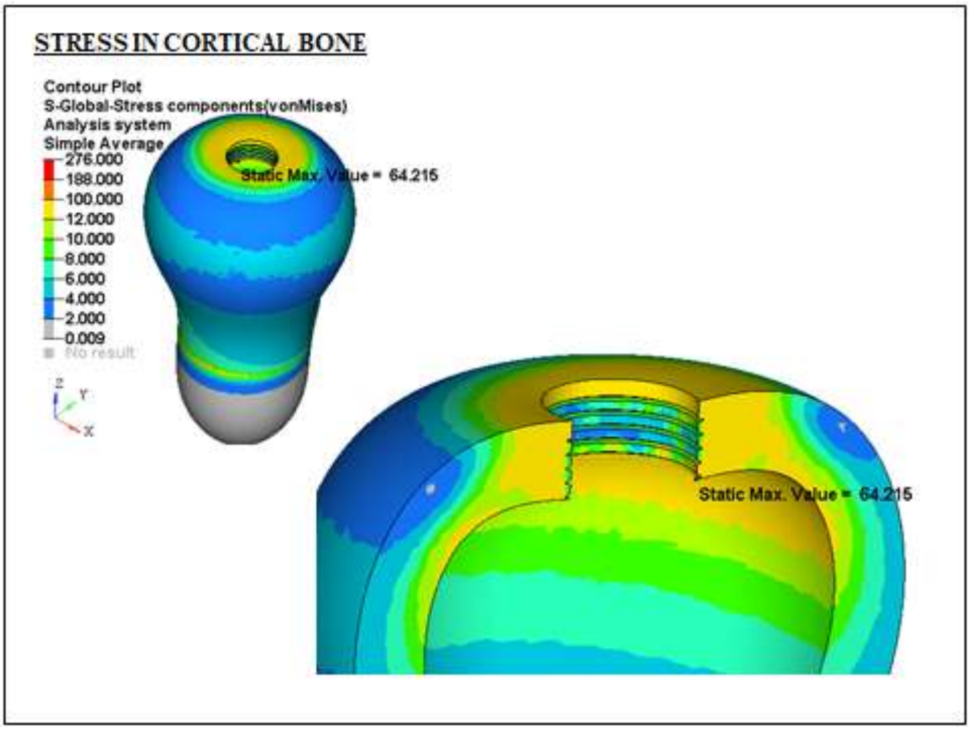


FIG. 15 : D1 LOADED AT 0 DEGREE

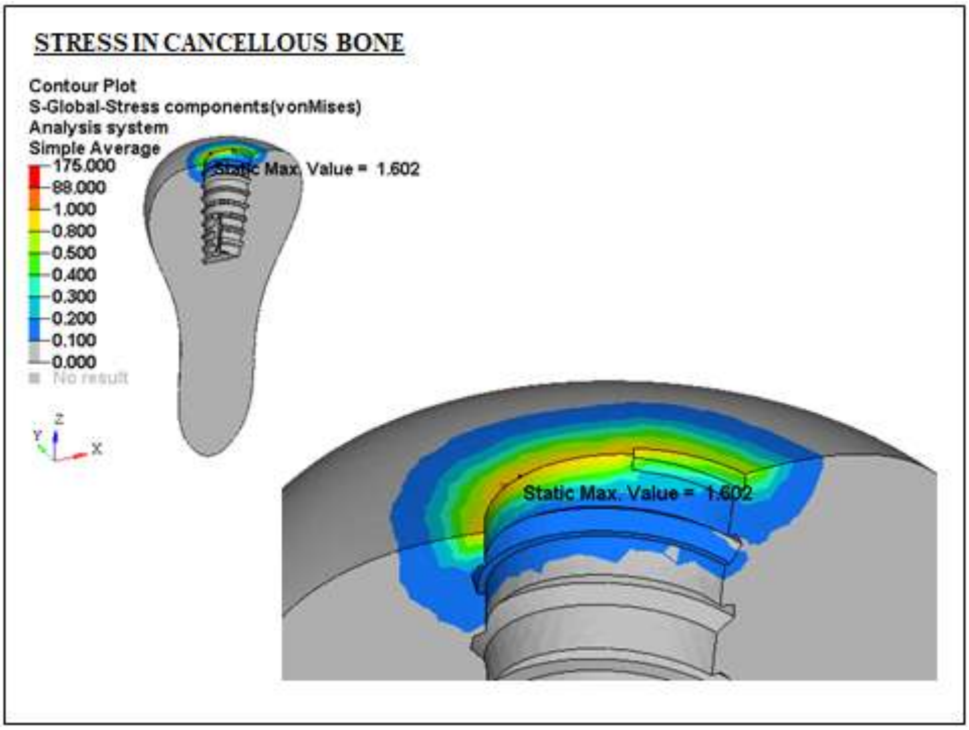


FIG. 16: D1 LOADED AT 5 DEGREE

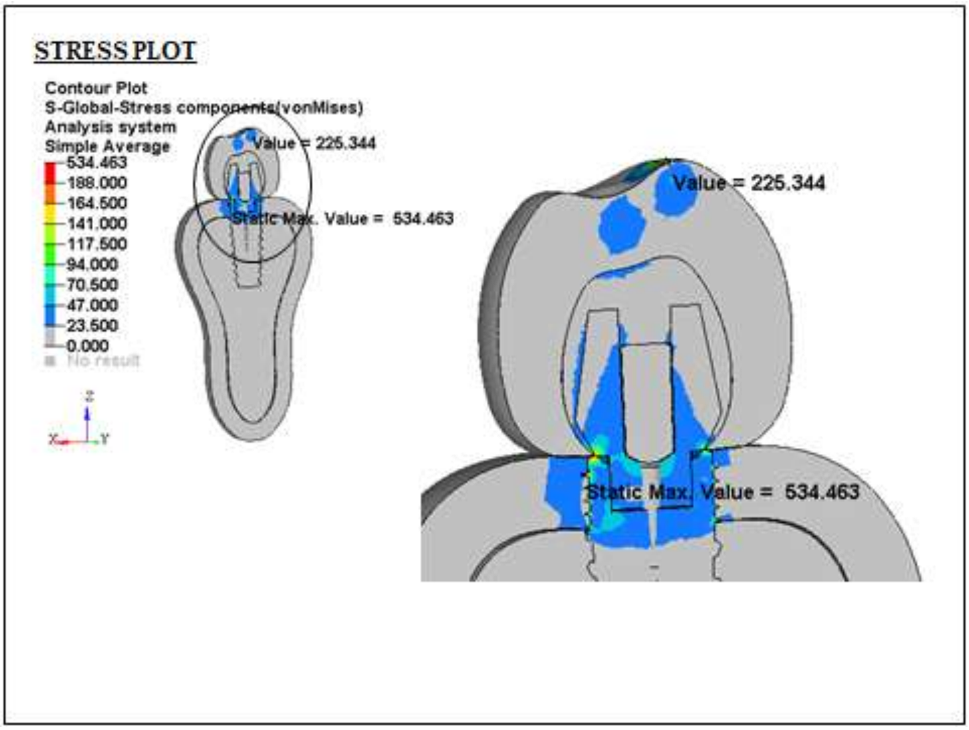


FIG. 17 : D1 LOADED AT 5 DEGREE

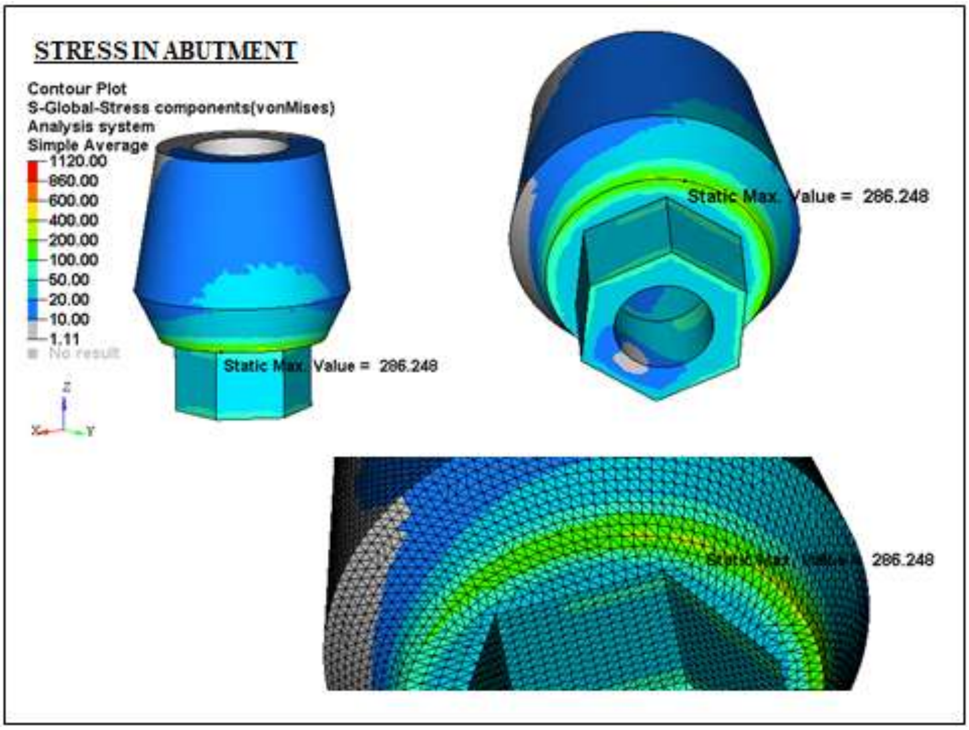


FIG. 18 D1 LOADED AT 5 DEGREE

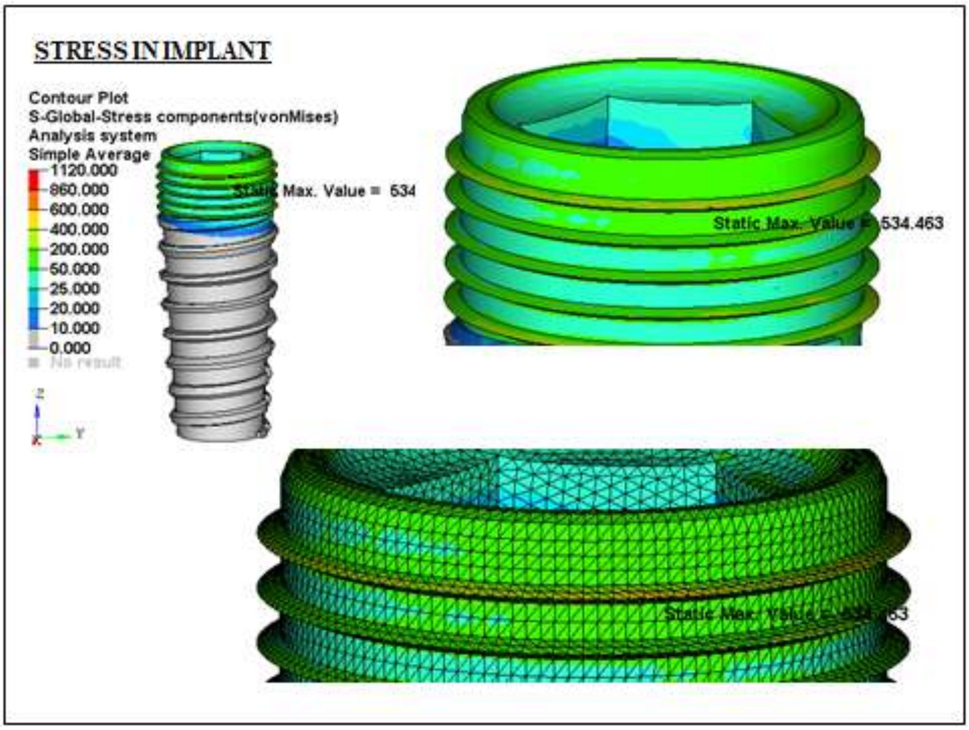


FIG. 19 : D1 LOADED AT 5 DEGREE

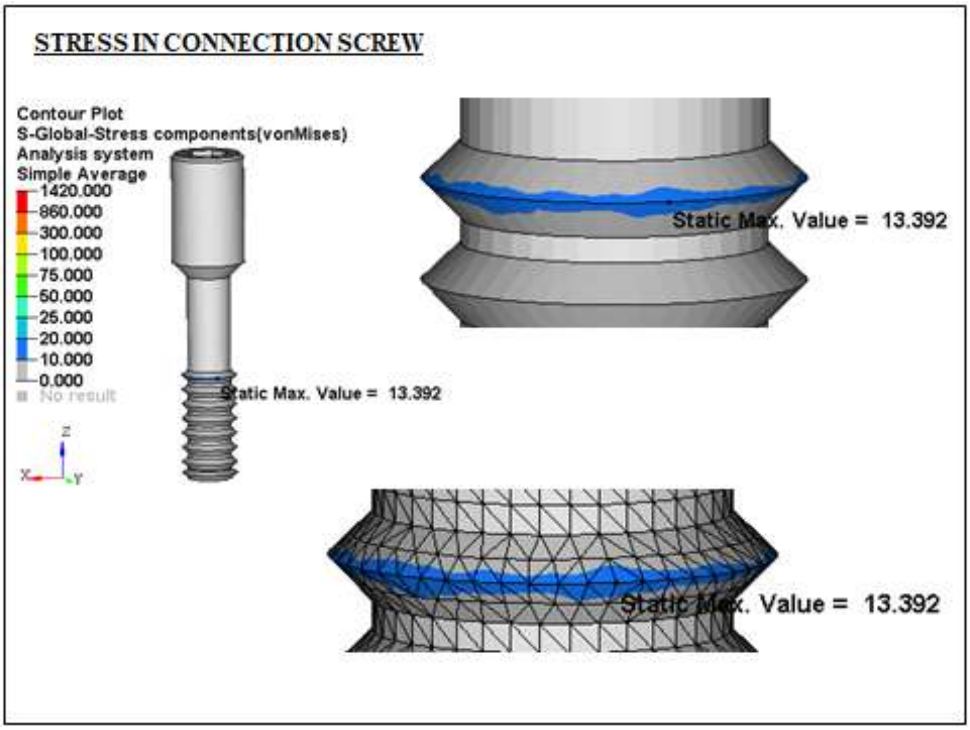


FIG. 20 : D1 LOADED AT 10 DEGREE

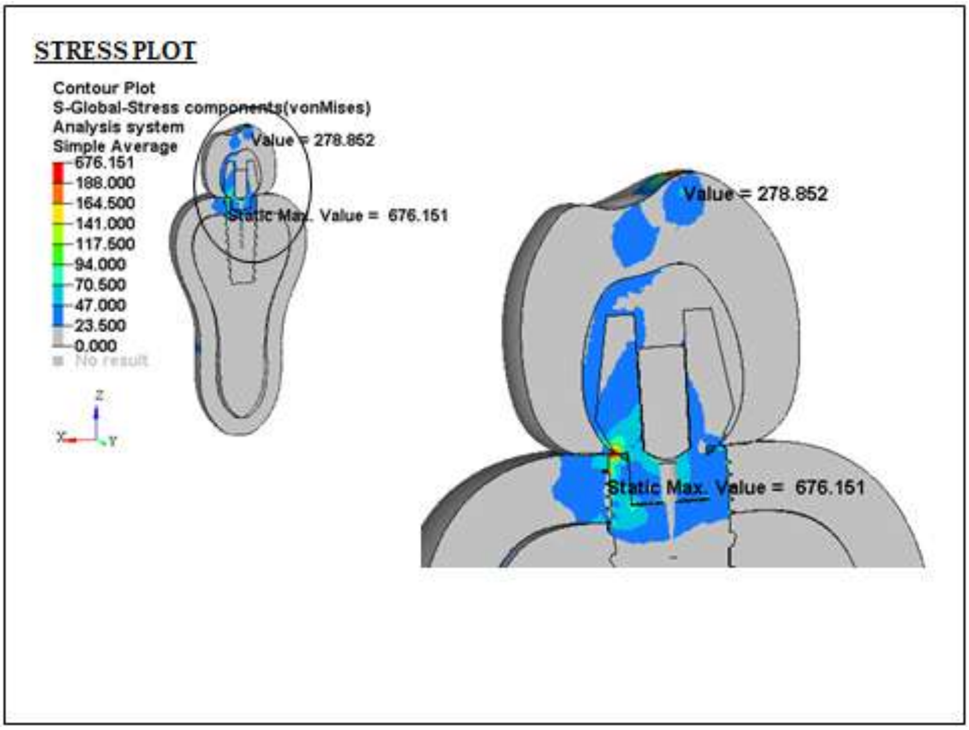


FIG.21: D1 LOADED AT 10 DEGREE

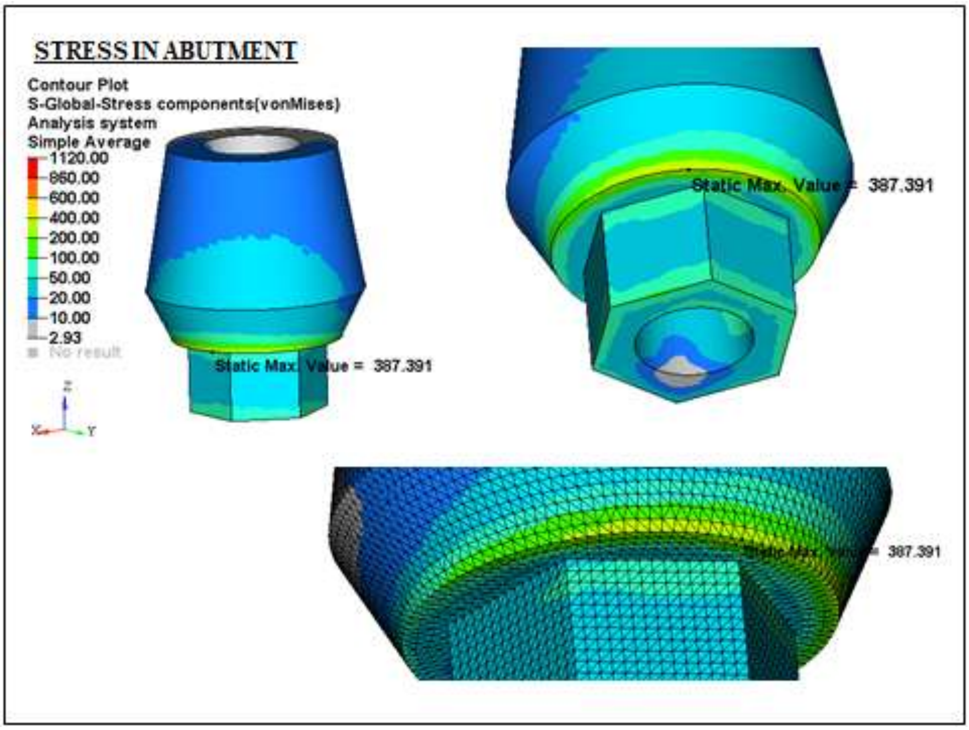


FIG. 22: D1 LOADED AT 10 DEGREES

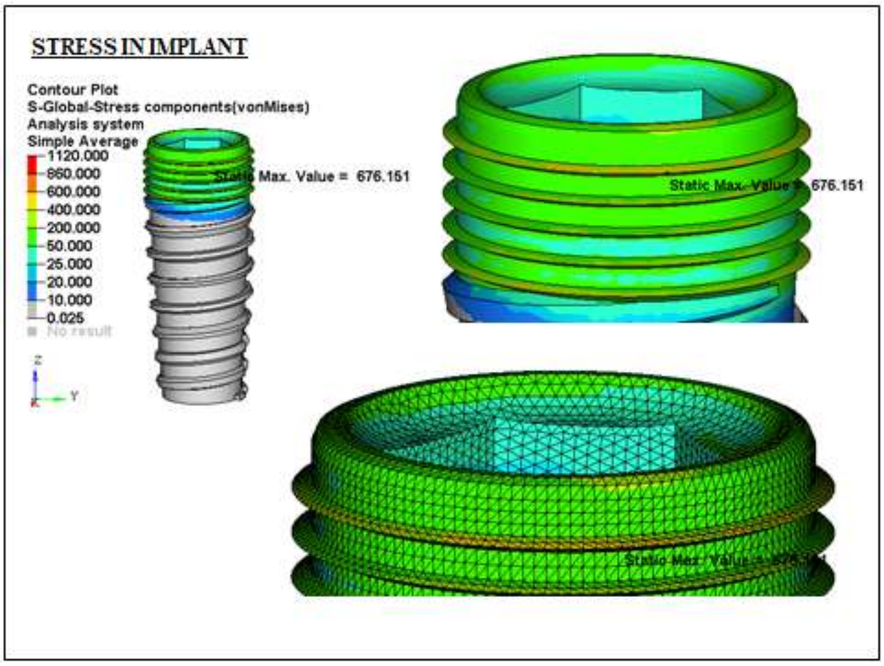


FIG. 23: D1 LOADED AT 10 DEGREE

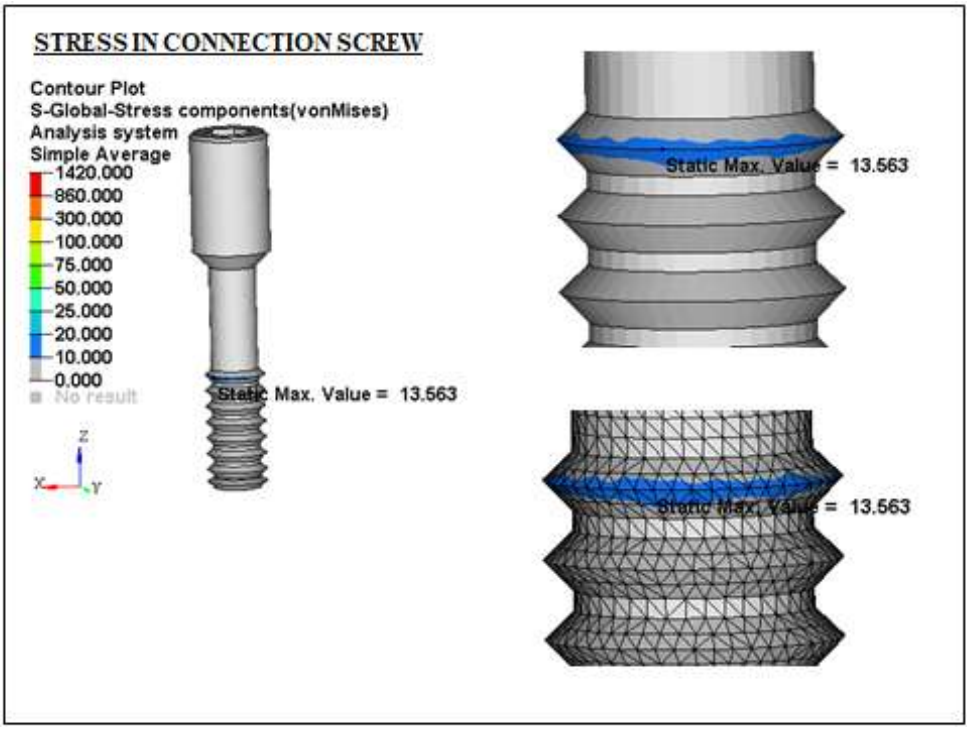


FIG. 24 D1 LOADED AT 15 DEGREE

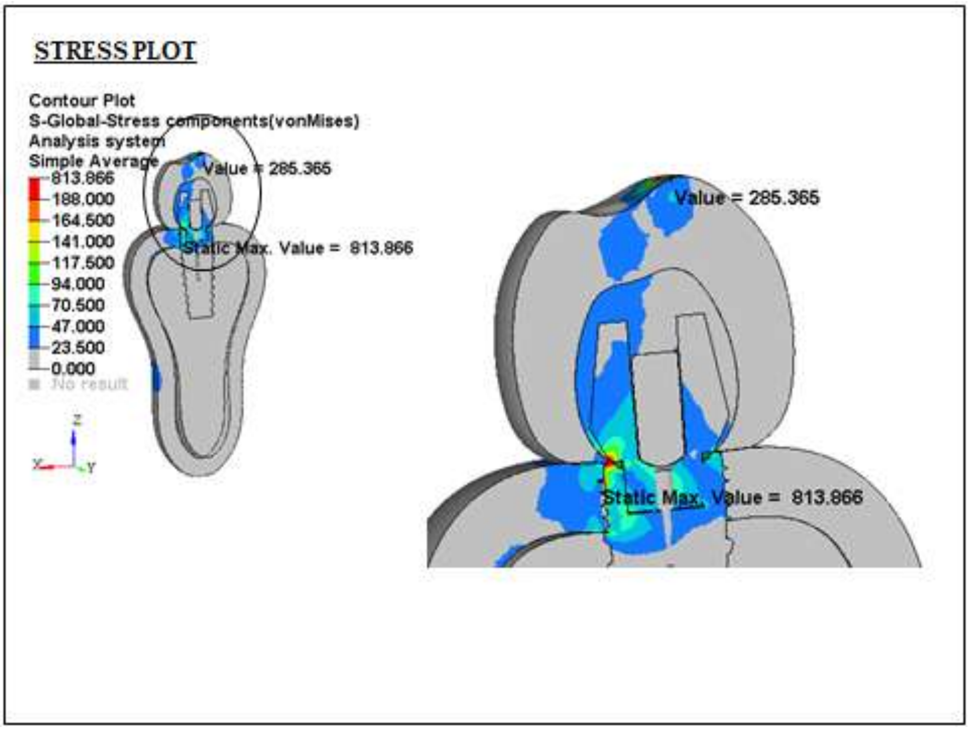


FIG. 25 : D1 LOADED AT 15 DEGREE

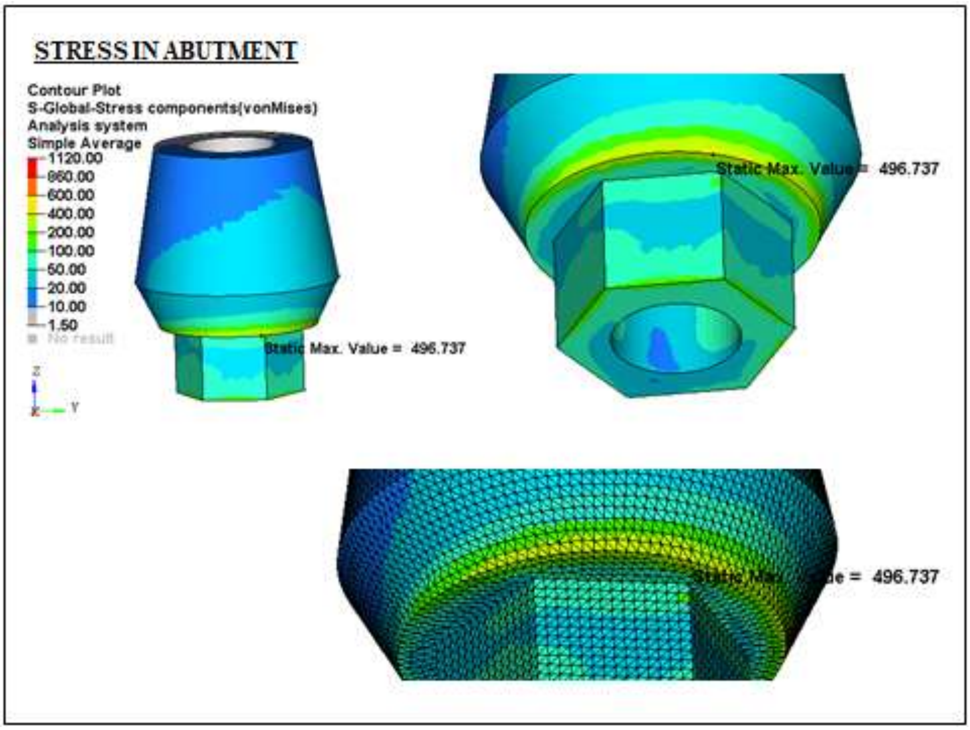


FIG. 26: D2 LOADED AT 0 DEGREE

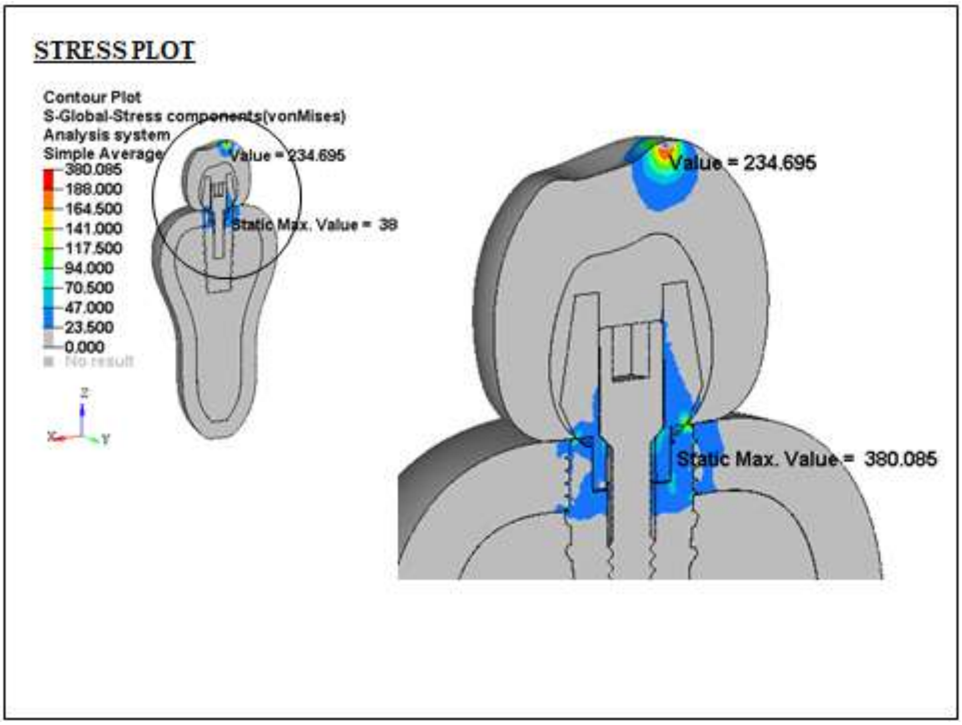


FIG. 27 : D2 LOADED AT 0 DEGREE

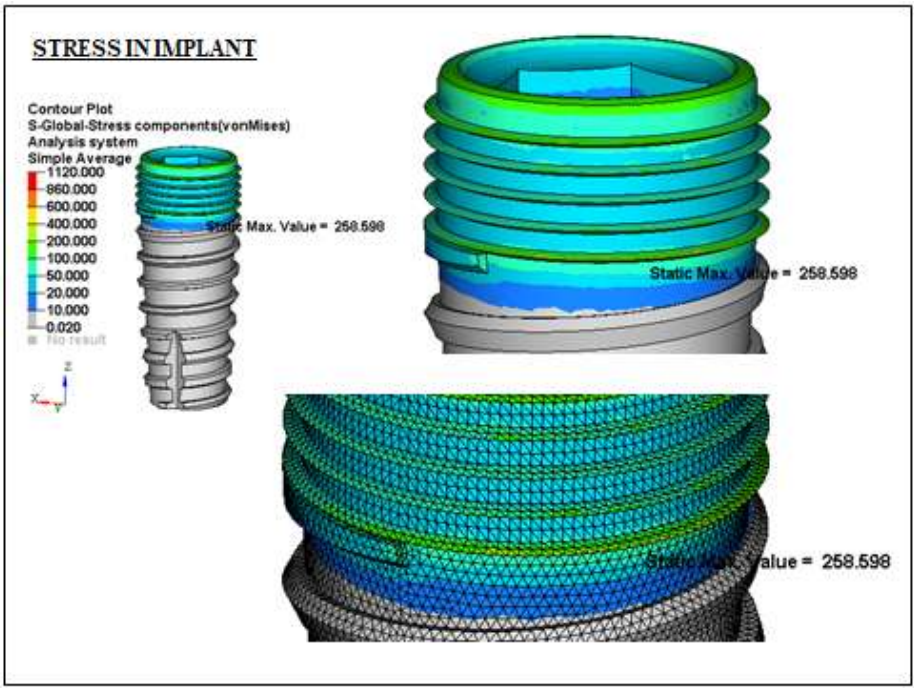


FIG. 28: D3 LOADED AT 0 DEGREE

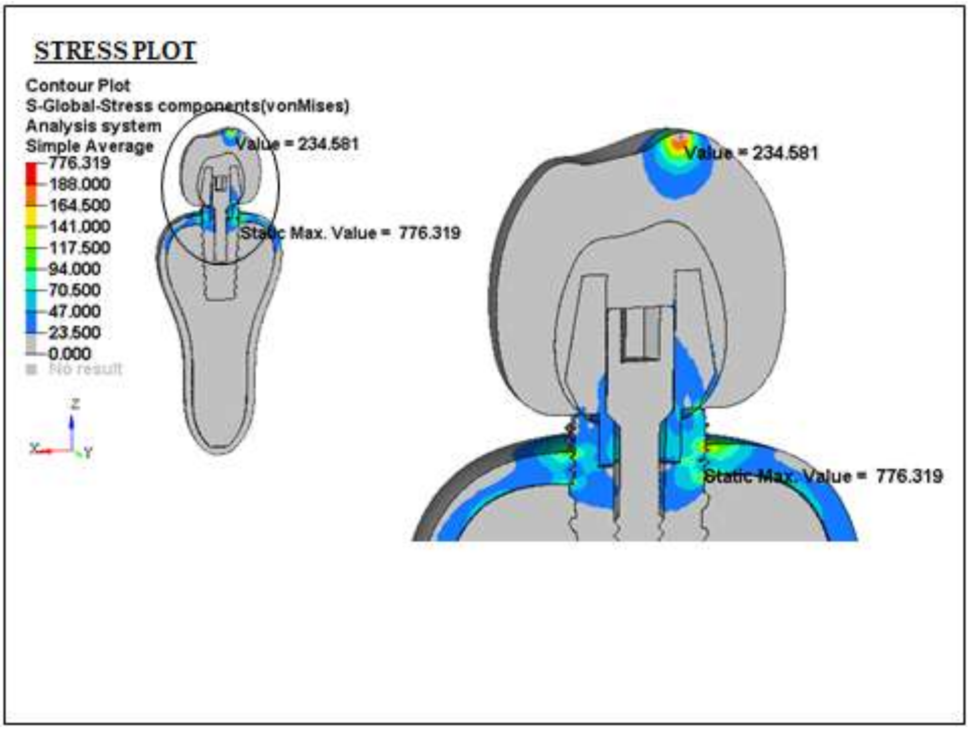


FIG. 29: D3 LOADED AT 0 DEGREE

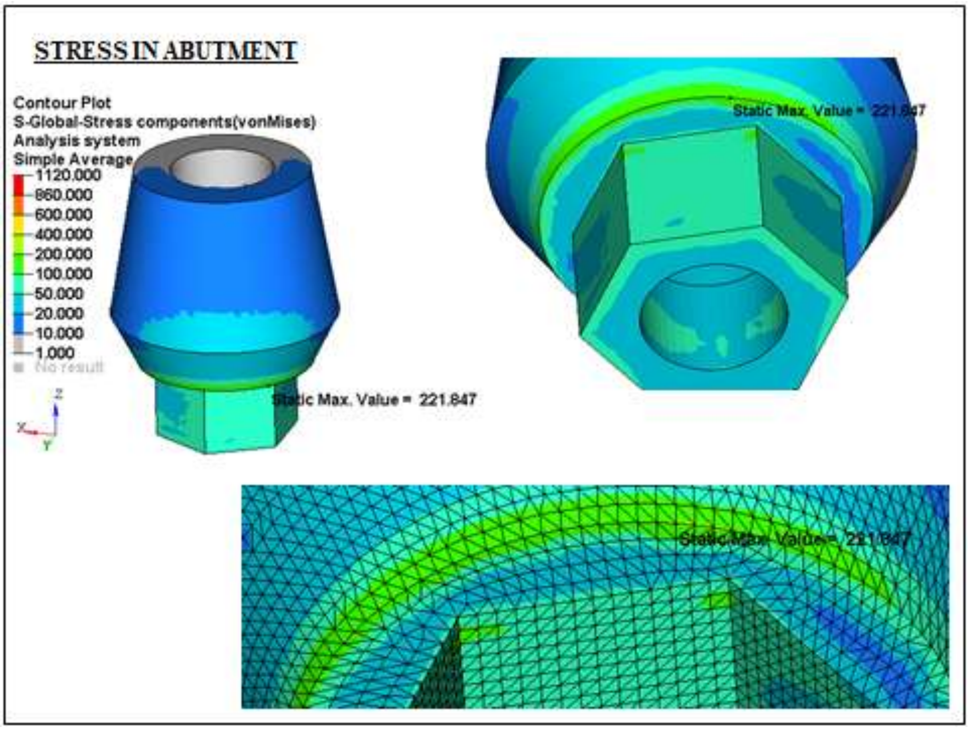


FIG.30 D4 LOADED AT 0 DEGREE

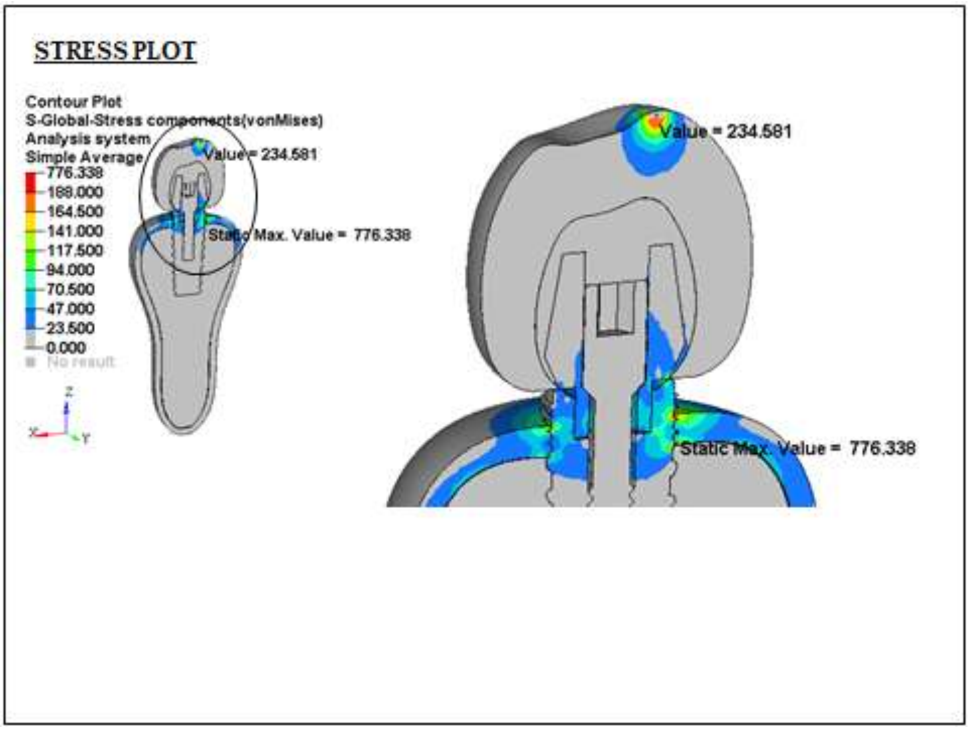
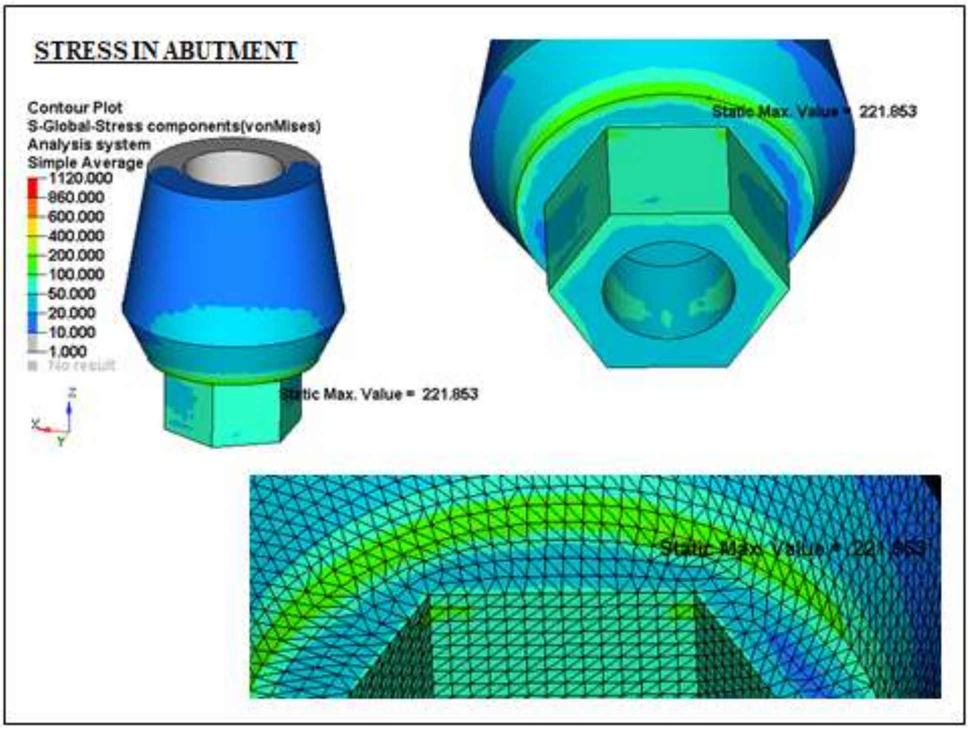


FIG. 31 : D4 LOADED AT 0 DEGREE



CERTIFICATE - II

This is to certify that this dissertation work titled ……………………………….. of the candidate ……………………………………… with registration Number ……………for the award of …………………………………………. in the branch of ………………….. . I personally verified the urkund.com website for the purpose of plagiarism Check. I found that the uploaded thesis file contains from introduction to conclusion pages and result shows ……… percentage of plagiarism in the dissertation.

Guide & Supervisor sign with Seal.

ANOVA TEST

TABLE 1: Comparison of von mises stresses of Abutments in D1 D2 D3 and D4 bone models subjected to forces at different angulations

Angle of Load	Von misses stress (Mean ±SD)				P Value
	D1 (MPa)	D2 (MPa)	D3 (MPa)	D4 (MPa)	
0°	282.88±0.531	286.44±0.217	221.58±0.362	222.00±0.686	.0001
5°	286.27±0.121	289.17±0.035	218.22±0.131	218.23±0.407	
10°	387.12±0.535	388.15±0.496	308.53±0.254	308.42±0.524	
15°	496.97±0.435	499.27±0.175	399.21±0.098	399.20±0.105	

D - BONE DENSITY ; ° DEGREE ; MPa -MEGAPASCAL

TABLE 2: Comparison of von mises stresses of Connecting screw in D1 D2 D3 and D4 bone models subjected to forces at different angulations

Angle of Load	Von misses stress (Mean ±SD)				P Value
	D1 (MPa)	D2 (MPa)	D3 (MPa)	D4 (MPa)	
0°	131.247±0.475	122.200±0.519	207.720±0.672	209.372±0.468	.0001
5°	130.510±0.631	122.815±0.343	215.845±0.166	214.415±0.260	
10°	135.782±0.067	134.768±0.190	239.950±0.330	233.022±0.433	
15°	137.940±0.604	141.618±0.055	263.178±0.317	262.280±0.175	

D - BONE DENSITY ; ° DEGREE ; MPa -MEGAPASCAL

TABLE 3: Comparison of von mises stresses of Implants in D1 D2 D3 and D4 bone models subjected to forces at different angulations

Angle of Load	Von misses stress (Mean ±SD)				P Value
	D1 (MPa)	D2 (MPa)	D3 (MPa)	D4 (MPa)	
0°	539.35±0.308	258.41±0.230	776.56±0.287	775.63±0.498	.0001
5°	535.23±0.511	251.37±0.114	861.25±0.590	861.84±0.660	
10°	675.76±0.482	295.29±0.786	958.41±0.687	959.31±0.259	
15°	813.80±0.128	362.68±0.525	1018. ±0.205	1181.8±0.389	

D - BONE DENSITY ; ° DEGREE ; MPa -MEGAPASCAL

TABLE 4: Comparison of von mises stresses of Co-Cr crowns in D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of Load	Von misses stress (Mean ±SD)				P Value
	D1 (MPa)	D2(MPa)	D3 (MPa)	D4 (MPa)	
0°	379.49±0.477	380.93±0.899	372.37±0.336	371.84±0.477	.0001
5°	391.94±0.194	392.14±0.533	390.34±0.097	390.13±0.464	
10°	499.71±0.497	498.20±0.180	493.00±0.682	492.95±1.206	
15°	675.27±1.100	672.57±0.318	651.85±0.854	650.92±0.405	

D - BONE DENSITY ; ° DEGREE ; MPa -MEGAPASCAL

TABLE 5: Comparison of von mises stresses in Feldspathic porcelain of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of Load	Von misses stress (Mean ±SD)				P Value
	D1 (MPa)	D2 (MPa)	D3 (MPa)	D4 (MPa)	
0°	234.66±0.044	234.93±0.391	234.46±0.244	235.24±0.500	.0001
5°	225.79±1.381	226.30±0.895	225.64±0.491	224.28±0.777	
10°	281.27±0.135	230.47±0.161	231.21±0.488	230.91±0.474	
15°	250.07±0.747	249.46±0.516	250.04±0.480	249.18±0.712	

D - BONE DENSITY ; ° DEGREE ; MPa -MEGAPASCAL

TABLE 6: Comparison of von mises stresses in Cortical bone of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of Load	Von misses stress (Mean ±SD)				P Value
	D1 (MPa)	D2 (MPa)	D3 (MPa)	D4 (MPa)	
0°	63.931±0.541	59.312±0.491	179.56±0.372	179.02±0.648	.0001
5°	67.403±0.464	62.239±0.086	192.83±0.839	192.45±0.462	
10°	75.292±0.197	78.289±0.563	237.80±0.546	236.97±0.752	
15°	91.077±0.250	93.954±0.674	295.07±0.172	295.20±0.097	

D - BONE DENSITY ; ° DEGREE ; MPa -MEGAPASCAL

TABLE 7: Comparison of von mises stresses in Cancellous bone of D1 D2 D3 and D4 models subjected to forces at different angulation

Angle of Load	Von misses stress (Mean ±SD)				P Value
	D1 (MPa)	D2 (MPa)	D3 (MPa)	D4 (MPa)	
0°	1.592±0.187	1.523±0.200	2.774±0.573	2.130±0.079	.0001
5°	1.469±0.104	1.351±0.118	3.237±0.316	2.446±0.285	
10°	1.736±0.050	1.700±0.045	3.08±0.273	2.574±0.289	
15°	2.266±0.197	1.557±0.443	4.308±0.310	3.350±0.001	

D - BONE DENSITY ; ° DEGREE ; MPa -MEGAPASCAL

TUCKEY POST HOC TEST:

TABLE 8: Multiple comparison of von mises stresses in abutments of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of load	Group comparison (p value)					
	D1 vs D2	D1 vs D3	D1 vs D4	D2 vs D3	D2 vs D4	D3 vs D4
0°	0.0001	0.0001	0.0001	0.0001	0.0001	0.618
5°	0.0001	0.0001	0.0001	0.0001	0.0001	1.000
10°	0.038	0.0001	0.0001	0.0001	0.0001	0.987
15°	0.0001	0.0001	0.0001	0.0001	0.0001	1.000

TABLE 9: Multiple comparison of von mises stresses in connecting screw of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of load	Group comparison (p value)					
	D1 vs D2	D1vs D3	D1 vs D4	D2 vs D3	D2 vs D4	D3 vs D4
0°	0.137	0.0001	0.0001	0.0001	0.0001	0.972
5°	0.065	0.0001	0.0001	0.0001	0.0001	0.952
10°	0.959	0.0001	0.0001	0.0001	0.0001	0.025
15°	0.483	0.0001	0.0001	0.0001	0.0001	0.983

TABLE 10: Multiple comparison of von mises stresses in implants of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of load	Group comparison (p value)					
	D1 vs D2	D1vs D3	D1 vs D4	D2 vs D3	D2 vs D4	D3 vs D4
0°	0.0001	0.0001	0.0001	0.0001	0.0001	.011
5°	0.0001	0.0001	0.0001	0.0001	0.0001	0.414
10°	0.0001	0.0001	0.0001	0.0001	0.0001	0.194
15°	0.0001	0.0001	0.0001	0.0001	0.0001	0.719

TABLE 11: Multiple comparison of von mises stresses in Co-Cr crown of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of load	Group comparison (p value)					
	D1 vs D2	D1vs D3	D1 vs D4	D2 vs D3	D2 vs D4	D3 vs D4
0°	0.021	0.0001	0.0001	0.0001	0.0001	0.592
5°	0.865	0.0001	0.0001	0.0001	0.0001	0.850
10°	0.059	0.0001	0.0001	0.0001	0.0001	1.000
15°	0.001	0.0001	0.0001	0.0001	0.0001	0.336

TABLE 12: Multiple comparison of von mises stresses in feldspathic porcelain of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of load	Group comparison (p value)					
	D1 vs D2	D1vs D3	D1 vs D4	D2 vs D3	D2 vs D4	D3 vs D4
0°	0.692	0.829	0.133	0.256	0.592	0.31
5°	0.869	0.995	0.158	0.755	0.044	0.224
10°	0.0001	0.0001	0.0001	0.054	0.354	0.639
15°	0.542	1.000	0.238	0.574	0.916	0.258

TABLE 13: Multiple comparison of von mises stresses in cortical bone of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of load	Group comparison (p value)					
	D1 vs D2	D1vs D3	D1 vs D4	D2 vs D3	D2 vs D4	D3 vs D4
0°	0.0001	0.0001	0.0001	0.0001	0.0001	0.496
5°	0.0001	0.0001	0.0001	0.0001	0.0001	0.754
10°	0.0001	0.0001	0.0001	0.0001	0.0001	0.202
15°	0.0001	0.0001	0.0001	0.0001	0.0001	0.956

TABLE 14: Multiple comparison of von mises stresses in cancellous bone of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of load	Group comparison (p value)					
	D1 vs D2	D1vs D3	D1 vs D4	D2 vs D3	D2 vs D4	D3 vs D4
0°	0.990	0.001	0.136	0.001	0.083	0.062
5°	0.882	0.0001	0.0001	0.0001	0.0001	0.002
10°	0.994	0.0001	0.0001	0.0001	0.0001	0.016
15°	0.020	0.0001	0.001	0.0001	0.0001	0.002

ANOVA:

TABLE 15: Percentage difference of Von mises stresses in abutments of D1 D2 D3 D4 models loaded at 0°, 5°, 10°and 15°

Percentage difference	Bone density	N	Mean (%)	Standard deviation
0° - 15°	D1	4	43.078	0.154
	D2		42.628	0.036
	D3		44.494	0.099
	D4		43.647	0.185
0° - 5°	D1	4	1.184	0.224
	D2		0.945	0.069
	D3		-1.540	0.108
	D4		-1.728	0.153
5° - 10°	D1	4	0.127	0.127
	D2		0.092	0.092
	D3		0.057	0.057
	D4		0.227	0.227
10 - 15°	D1	4	22.103	0.175
	D2		22.255	0.126
	D3		22.715	0.051
	D4		22.742	0.117

TABLE 16: Multiple comparisons for percentage difference of stresses in Abutments of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of load	Group Comparison (p value)					
	D1 vs D2	D1vs D3	D1 vs D4	D2 vs D3	D2 vs D4	D3 vs D4
0° - 5°	0.166	0.001	0.001	0.001	0.001	0.332
5° - 10°	0.001	0.001	0.001	0.001	0.001	0.991
10° - 15°	0.361	0.001	0.001	0.001	0.001	0.990
0° - 15°	0.002	0.001	0.001	0.001	0.001	0.675

ANOVA:

TABLE 17: Percentage difference of Von mises stresses in Implant body of D1 D2 D3 D4 models loaded at 0°, 5°, 10°and 15°

Percentage Difference	Bone density	N	Mean (%)	Standard deviation
0° - 5°	D1	4	-0.769	0.054
	D2		-2.801	0.110
	D3		9.833	0.091
	D4		10.002	0.086
5° - 10°	D1	4	20.796	0.116
	D2		14.874	0.248
	D3		10.140	0.037
	D4		10.160	0.052
10° - 15°	D1	4	16.962	0.048
	D2		18.579	0.130
	D3		18.881	0.043
	D4		18.825	0.016
0 - 15°	D1	4	33.725	0.044
	D2		28.749	0.043
	D3		34.274	0.035
	D4		34.368	0.046

TABLE 18: Multiple comparisons for percentage difference of stresses in implant body of D1 D2 D3 and D4 models subjected to forces at different angulations

Angle of load	Group comparison (p value)					
	D1 vs D2	D1vs D3	D1 vs D4	D2 vs D3	D2 vs D4	D3 vs D4
0° - 5°	0.001	0.001	0.001	0.001	0.001	0.076
5° - 10°	0.001	0.001	0.001	0.001	0.001	0.997
10° - 15°	0.001	0.001	0.001	0.001	0.002	0.714
0° - 15°	0.001	0.001	0.001	0.001	0.001	0.400

RESULTS

The master chart of the results is given in the annexure.

The data collected from the study was further analysed with SPSS software (statistical package for social science) 16.0 version. ANOVA (analysis of variance) test was used to analyse the difference between the means of D1, D2, D3, D4 groups and percentage difference between the groups loaded at different angulations in the abutments and implant body category and for comparison within groups. While Tukey's post hoc test was done for multiple comparisons between the groups.

Calculation Tables:

Table 1 depicts the comparison of mean values for the groups D1, D2, D3, D4 in 0°, 5°, 10° and 15° for abutment . The D3 model loaded at 0 degrees shows the lowest mean stress values of 218.22±0.131 and for the D1 model at 15° the mean is 496.97±0.435, which is a gradual increase from 0° to 15°. And the highest values were recorded for the D2 model at 15° which is 499.27±0.175 angulation.

Table 2 depicts the comparison of mean values for the groups D1, D2, D3, D4 in 0°,5°,10°and 15° for the connecting screw. The D1 model loaded at 0 degrees shows the lowest mean stress values of 131.247±0.475 and when loaded at 15° it shows a higher value of 137.940±0.604 and D4 when loaded at 0° shows 209.372±0.468. And the highest value of mean for the whole table is recorded at 15° in D3, 263.178±0.317 closely followed by D4 which is 262.280±0.175 all of which shows significant P values of (0.0001).

Table 3 depicts the comparison of mean values for the groups D1, D2, D3, D4 in 0°,5°,10°,15° for the Implant. The D2 model loaded at 0° shows the lowest mean stress value of 258.41±0.230 and the highest mean stress values are in D4 at 0° which is 775.63±0.498 and at 15° it is 1181.8±0.38.

Table 4 depicts the comparison of mean values in the Co – Cr crown in groups D1, D2, D3, D4 at 0 °,5°,10°,15°. The lowest mean values were recorded in D4 at 0° as 371.84±0.477 and the highest mean stress values were recorded for D1 at 15° as 675.27±1.100 and all the values had significant P values.

Table 5 depicts the comparison of mean von misses stress values in groups D1, D2, D3, D4 at 0 °,5°,10°,15° in Feldspathic porcelain. The lowest mean values were recorded in D4 at 5° as 224.28±0.777 and the mean values recorded at 10° in D1 was 281.27±0.135 and all the values in the table had significant P values of 0.0001.

Table 6 depicts the comparison of mean von mises stress values in groups D1, D2, D3, D4 at 0 °,5°,10°,15° in cortical bone. The lowest mean values were recorded in D2 at 0 degrees as 59.312±0.491 and the highest stresses were recorded in D4 at 15° as 295.20±0.097.

Table 7 depicts the comparison of mean stresses in D1, D2, D3, D4 at 0 °,5°,10°,15° in cancellous bone. The lowest mean values were recorded in D2 at 5° as 1.351±0.118, and the highest mean stress values were recorded for D3 at 15°. And all the values were statistically significant.

Table 8 depicts the multiple inter group comparison of mean stresses by Tuckey‘s post hoc test in D1, D2, D3, D4 at 0°,5°,10°,15° in abutments which was highly significant (0.0001) for D1 vs D2, D1 vs D3, D1 vs D4, D2 vs D3, D2 vs D4.Whereas the D3 vs D4 shown P values as 0.6, 1, 0.9,1 for 0,5,10 and 15 degree angulations.

Table 9 depicts the multiple inter group comparison of mean stresses in D1, D2, D3, D4 at 0°,5°,10°,15° in the connecting screw which was highly significant (0.0001) for, D1 vs D3, D1 vs D4, D2 vs D3 ,D2 vs D4. Whereas the D3 vs D4 groups showed P values of 0.9, 0.9, 0.02, 0.9 and D1 vs D2 groups showed P values of 0.1, 0.06, 0.9, 0.4 for 0, 5, 10, 15 degrees.

Table 10 depicts the multiple inter group comparison of mean stresses in the implant category in D1, D2, D3, D4 at 0°,5°,10°,15° in implant which was highly significant (0.0001) in D1 vs D2, D1 vs D3, D1 vs D4, D2 vs D3, D2 vs D4 whereas the p values for D3 vs D4 were 0.01,0.4,0.1,0.7 in 0, 5,10, 15, degrees.

Table 11 depicts the multiple inter group comparison of mean stresses in D1, D2, D3, D4 at 0°,5°,10°,15° in Co – Cr crown which was highly significant (0.0001) in D1 vs D3, D1 vs D4, D2 vs D3,D2 vs D4. And P values of 0.02,0.8,0.05,0.001 in D1 vs D2 and 0.5,0.8,1,0.3 in D3 vs D4 groups.

Table 12 depicts the multiple inter group comparison of mean stresses in D1, D2, D3, D4 at 0°,5°,10°,15°in Feldspathic porcelain which was significant at 0.05 level for D1 vs D2, D1 vs D3, D1 vs D4, D2 vs D3, D2 vs D4, D3 vs D4 groups.

Table 13 depicts the multiple inter group comparison of mean stresses in D1, D2, D3, D4 at 0°,5°,10°,15° in cortical bone which was highly significant (0.0001) for D1 vs D2, D1 vs D3, D1 vs D4, D2 vs D3, D2 vs D4. And the P values were 0.4, 0.7, 0.2, 0.9 for D3 vs D4 groups.

Table 14 depicts the multiple inter group comparison of mean stresses in D1, D2, D3, D4 at 0°,5°,10°,15° in the cancellous bone which was significant in 0.05 level for D1 vs D2, D1 vs D3, D1 vs D4, D2 vs D3, D2 vs D4,D3 vs D4.

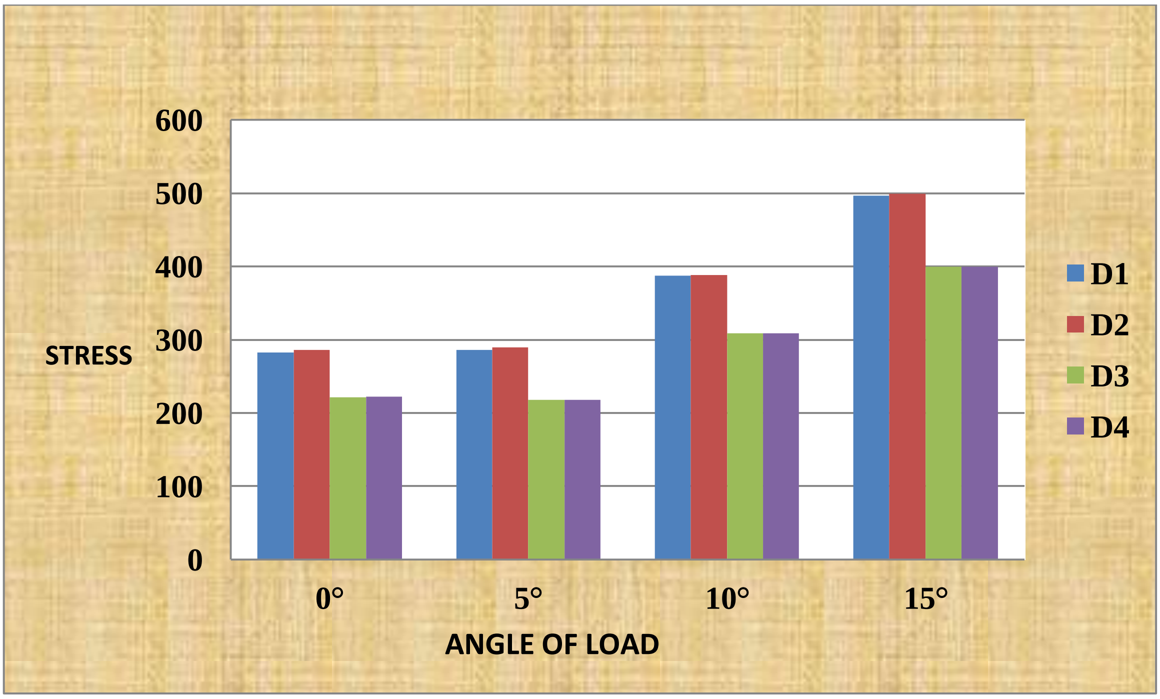
Table 15 depicts the percentage difference of von misses stresses in the abutment category by Anova test in D1, D2, D3, D4 groups at 0,5,10 and 15 degrees. With a mean percentage difference of 1% for D1 when loaded from 0° to 5° and a 0.1% for D1 when loaded from 5° to 10° and 22% for D1 when compared from 10° to 15°.And a 43% for D1 from 0° to 15° angulations.

Table 16 depicts multiple intergroup comparisons done by Tuckey post hoc test for percentage difference of stresses in abutments of D1, D2, D3, D4 groups at 0°,5°,10°,15°.And the P values were highly significant (0.001) for D1 vs D3, D1 vs D4, D2 vs D3, D2 vs D4. And in the D3 vs D4 the P values were 0.3, 0.9, 0.9, 0.6.

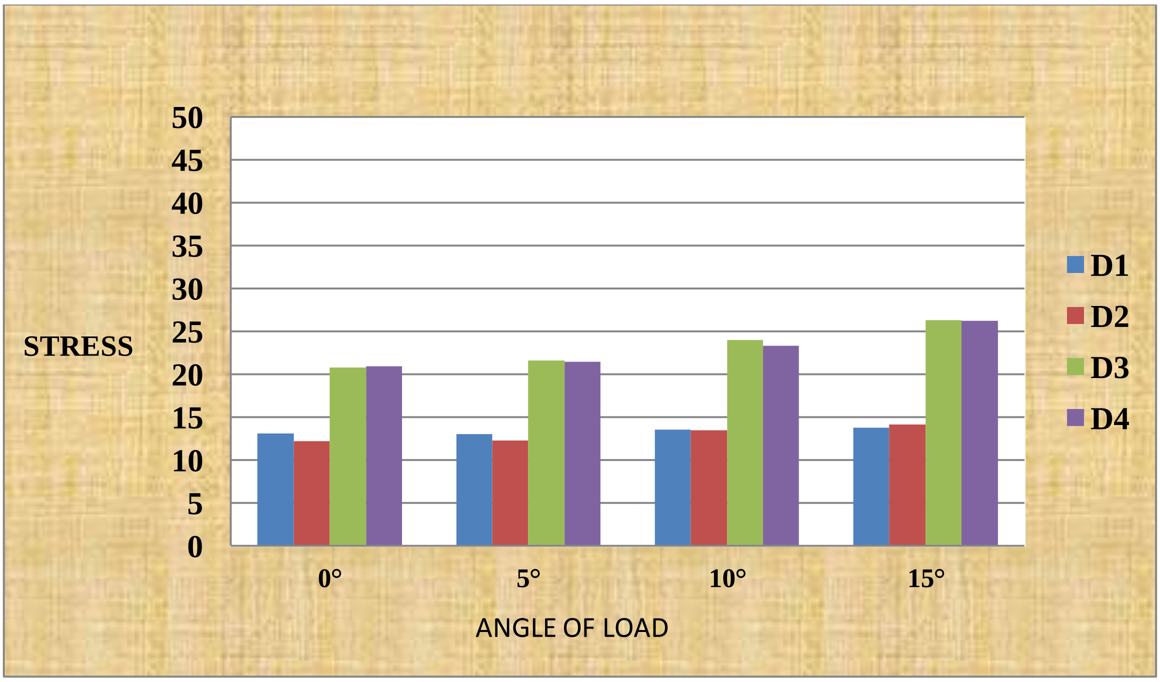
Table 17 depicts the percentage difference of von mises stresses in the implant body of D1, D2, D3, D4 models when compared between 0° - 5°, 5° - 10°, 10° - 15° and 0° - 15° models using ANOVA test. With a mean percentage difference of 0.7%(D1), 2.8%(D2), 9.8%(D3), 10%(D4) in 0° to 5° category, a mean percentage difference of 20% (D1), 14%(D2), 10%(D3), 10%(D4) in 5° -10° category, a mean percentage difference of 16%(D1), 18%(d2), 18%(D3), 18%(D4) in 10° - 15° category, a percentage difference of 33%(D1), 28%(D2), 34%(D3),34%(D4) in the 0° - 15° groups.

Table 18 depicts multiple intergroup comparisons by Tuckey's post hoc test for percentage difference of stresses in the implant body of D1, D2, D3, D4 models when compared between 0° - 5°, 5° - 10°, 10° - 15° and 0° - 15°. And the P values were highly significant (0.001) for D1 vs D3, D1 vs D4, D2 vs D3, D2 vs D4. And in the D3 vs D4 the P values were 0.07, 0.9, 0.7, 0.4.

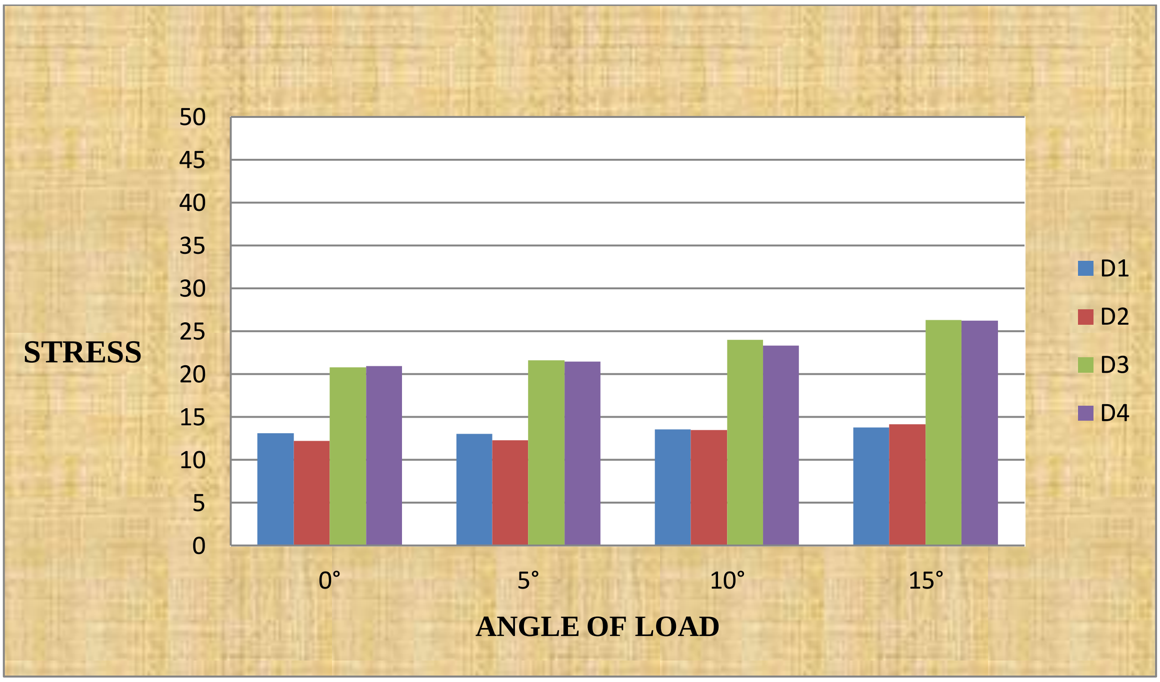
Graph 1: <u>Bar diagram Showing Von Mises Stresses In abutment for D1, D2, D3, D4</u>



Graph 2: <u>Bar diagram Showing Von Mises Stresses In Connecting Screw for D1, D2, D3, D4</u>

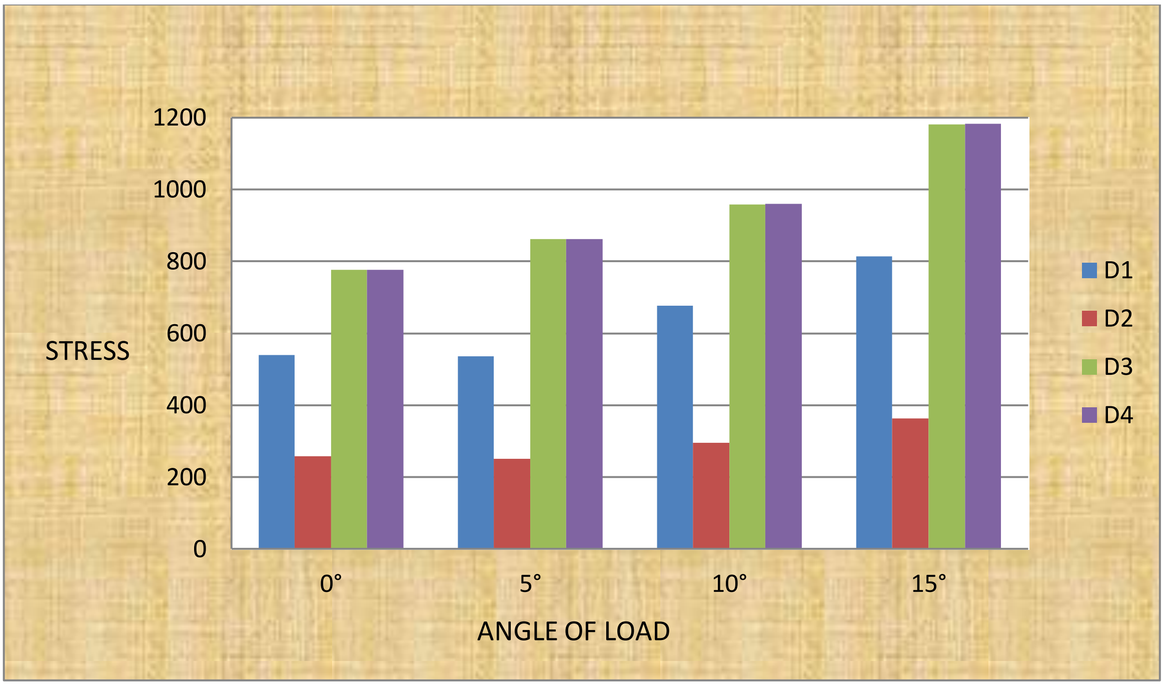


Graph 3: <u>Bar diagram showing von mises stresses in Implant for D1, D2, D3, D4</u>

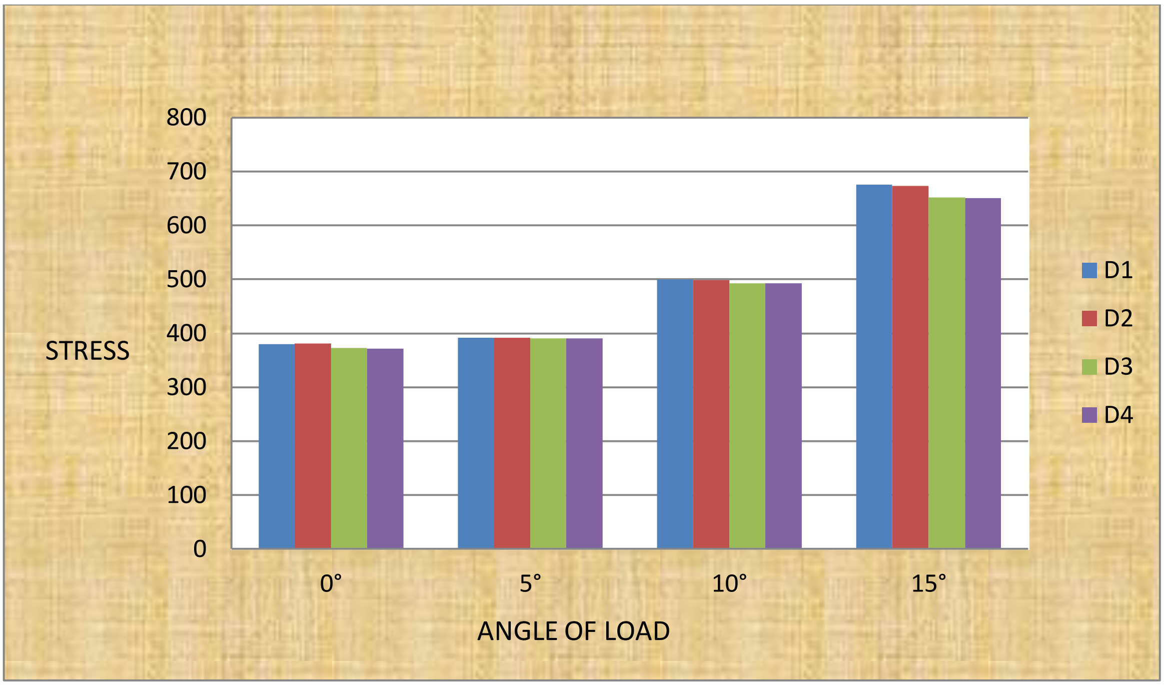


Graph 4: <u>Bar diagram showing von mises stresses in Co – Cr crown</u>

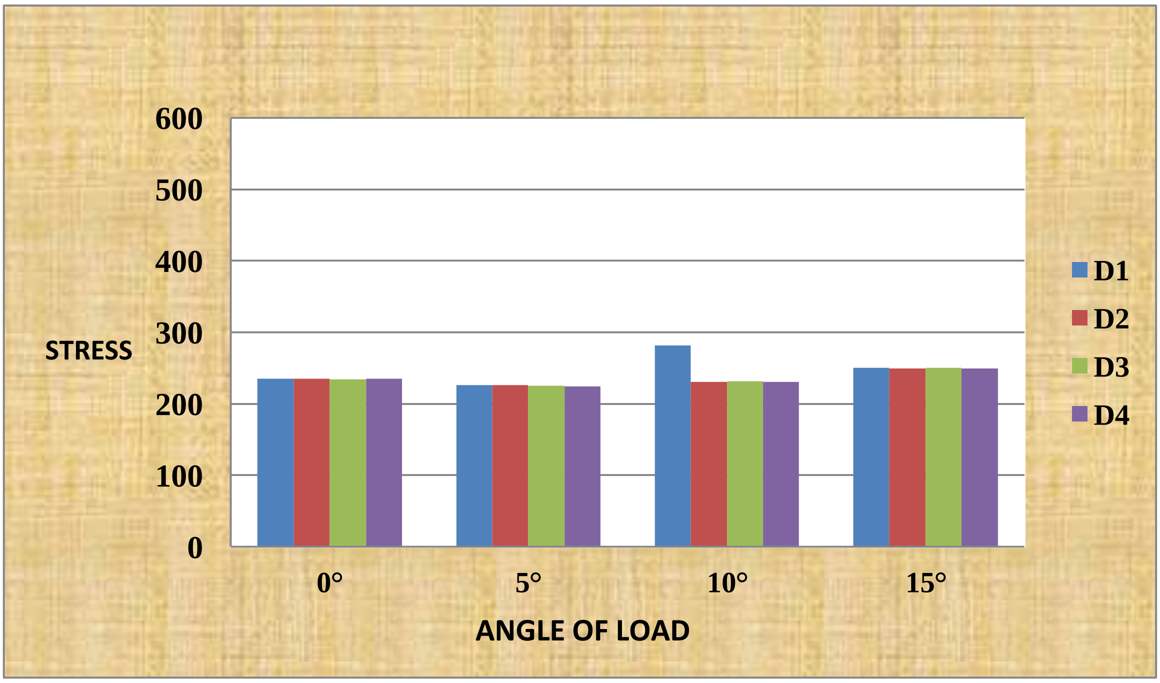
<u>for D1, D2, D3, D4</u>



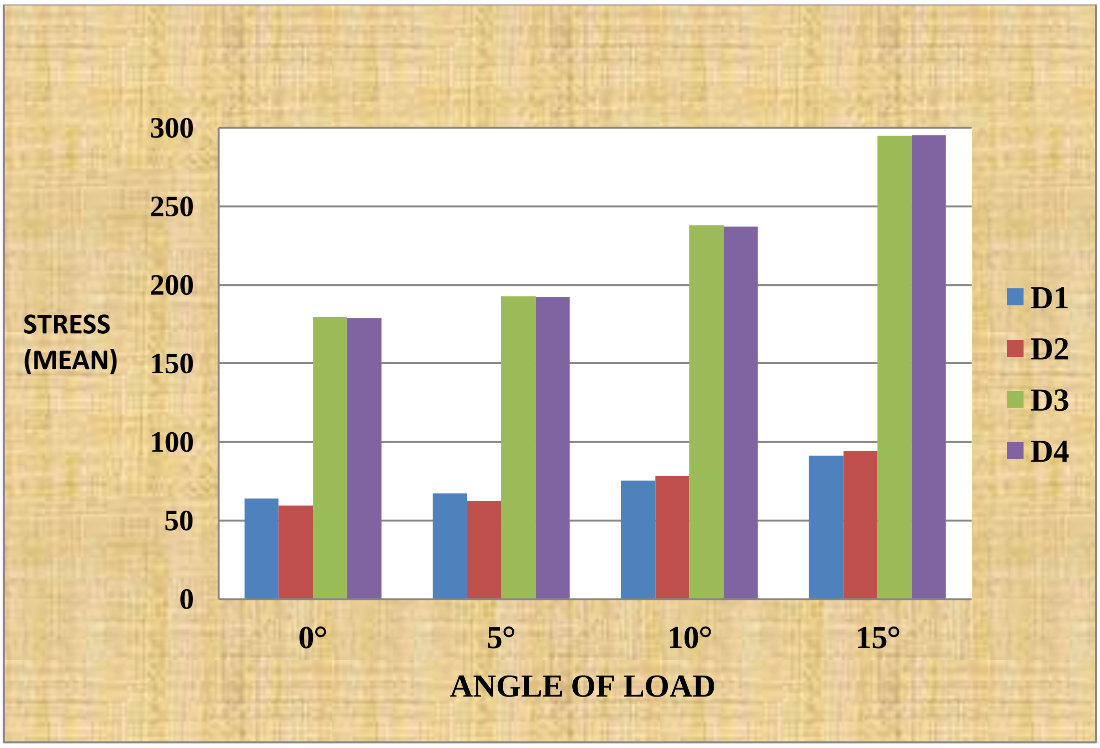
Graph 5: Bar diagram showing von mises stresses in Feldspathic porcrelain for D1, D2, D3, D4



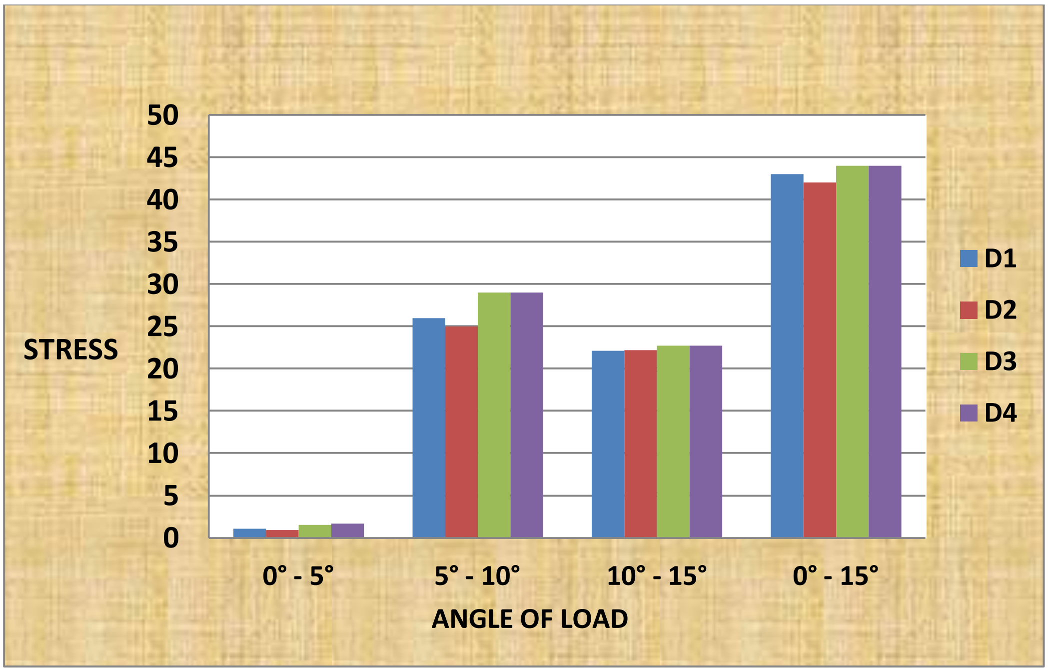
Graph 6: Bar diagram Showing Von Mises Stresses in Cortical bone for D1, D2, D3, D4



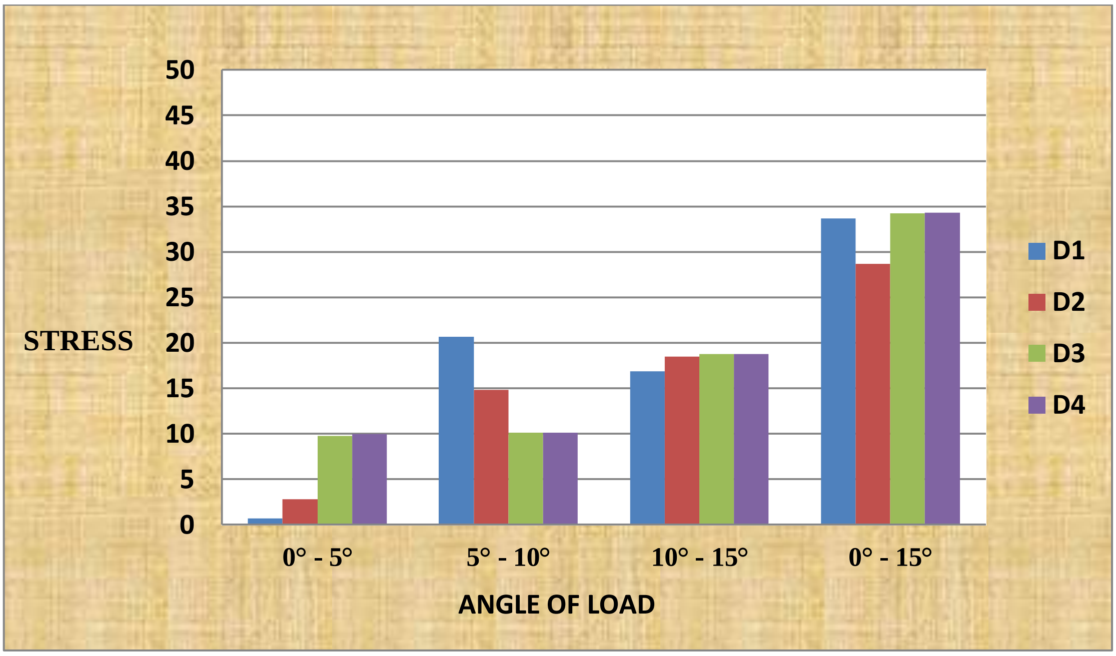
Graph 7: Bar diagram showing stresses in Cancellous bone for D1, D2, D3, D4



Graph 8: <u>Bar diagram showing percentage difference for stresses in Abutment for D1, D2, D3, D4 for 0°, 5°, 10° and 15°</u>



Graph 9: <u>Bar diagram showing percentage difference for von mises stresses in Implant body for D1, D2, D3, D4 for 0°, 5°, 10° and 15°</u>



DISCUSSION

Stress in a biologic environment is challenging, more so when there is a mismatch in relationship between the two materials in this instance modulus of elasticity of titanium (young's modulus for Ti is 10 times more than cortical bone and 20 times more for trabecular bone[60]) and bone and when the cellular biomechanics are involved additionally. While this may bring about an imbalance in the stress patterns which may lead to marginal bone loss eventually affecting the success of the implant restoration.

Biomechanics In Bone

STRESS: (internal force)

Stress = force/ area

The magnitude of stress depends on two variables,

a. Force magnitude

b. Cross sectional area over which the force is dissipated

so in order to minimize and equally distribute stress the above two dependent variables has to be regulated but it is not possible for dentist to control the force completely however the functional surface area can be controlled through careful treatment planning.

STRAIN: (external force)

Strain = change in length/ original length

Is the relative change in shape or size of an object due to an externally applied force.

Strain regulates the cellular activities of bone. When bone has ideal strain the bone remains organised and load bearing. When the strain is greater, it may be in a pathologic overload zone[61], which cause bone loss. Therefore it is hypothesised that occlusal stresses beyond the physiologic limit of bone may result in strain significant enough to cause bone resorption. After a specific implant system is decided, the only way to control strain on tissues is to control the applied stress.

Bone is a unique structure that is able to change in relation to a number of factors including hormones, vitamins, minerals and mechanical impact. However biomechanical parameters such as amount of strain transmitted to bone are predominant. In 1888, Kulman[62] identified similarities between the patterns of trabecular bone in the femur and tension trajectories in construction beam concepts used by Eiffel. In 1892, Wolfe[63] elaborated on the same concept and published an article where he states that bone is constantly modelled and remodelled according to the biomechanical needs. “Every change in form and function of bone or its function alone is followed by certain definite changes in the internal architecture and equally definitive alteration in its external conformation in accordance with mathematical laws.

In the present study, A total of 16 test runs was done on 4 CAD models in all four angulations with a load of 300 N[56] in order to evaluate the stress distribution patterns in 7 regions of interest within the model.

Loading criteria in the Study

The greatest natural forces acting on teeth are those that occur during mastication[64] and the forces applied are perpendicular to the occlusal plane in the posterior region. Apart from this a constant slight force is supplied by the perioral musculature and tongue. According to studies the normal maximum bite force ranges from 45 to 550 psi (200N - 2446N)[65]. The maximum bite force for the natural teeth or implants in the second premolar region is 583 ± 99 N (van eijden[56], 1991). Implants lack occlusal awareness so the velocity of the occlusal force before contact is not reduced. Hence the force factors in implant prosthesis are to be thoroughly evaluated so as deleterious stresses are not generated[66]. Hence in this study a 300 N bite force was assigned for loading the models in order to analyse the implant complex survival even during the higher end of occlusal forces but also well within the normal bite forces incurred in the premolar region.

Not only does the magnitude and duration of force plays a significant role in the survival of the prosthesis but equally important are the mechanical properties of the implant and the crown superstructure. Titanium as a biocompatible material is well documented[s1]. Titanium alloy, in comparison with grade 1, grade 2, grade 3, grade 4 Ti has the highest yield strength of 860MPa[67,768].And a tensile strength of 930 MPa which is greater than most of the restorative materials used in dentistry.

Density :

ρ = m/V ; where m – mass V - volume

Bone density is directly proportional to strength and elastic modulus of bone. In dense bone there is less stress under a given load when compared with softer bone types [69,70] The elastic modulus differs for all four qualities in a way that it is more for densest bone (D1 type), And the least for the poor bone types. And as the young's modulus increases the stiffness of the bone is increased wherein the strength of the bone is also increased and hence it is more ideal for load bearing.

Von misses stress distribution in different regions within the model:

In the present study the von mises stresses for the abutment, connecting screw and implant in all 4 implant types were least for the D1, D2 models and highest stresses are seen in the D3, D4 model. And among them the maximum stresses were found in the neck of the implant compared to other components.

Adell et al[21] conducted a 15 year follow up on 700 people who were regularly reviewed by annual clinical and radiographic methods. And the survival rates for maxillary implants were given for 5, 10 and 15 year time interval. And it was estimated that there was 78% survival at 15 years whereas in maxilla it was 99% at all time intervals throughout the study and states a 10% increased success rate in anterior mandible (dense bone) than in anterior maxilla.

Snauwaert et al[29] from his 15 year follow up study on 4971 implants in 1315 patients who were divided into two groups such as physical well and ill patient (maxilla and mandible) with 6 – 12 month recall period and implant failures were 11.6% in the non - compromised whereas in the compromised patients it was 40.6% and concluded that too they were more frequent in the maxilla.

Enguist et al[71] a retrospective study on implant supported over dentures which stated that there was more of failures in maxilla even before the loading of the prosthesis, while the mandibular overdentures showed a survival rate of 99% . And observed that 78% of all reported implant failures were in poor bone density types in overdentures.

Johns B jr[72] et al along with **Torsten Jemt, Robin Heath MR,** conducted a multicentre study on 133 patients with 117 implants in maxilla and 393 implants in the mandible and researches reports that most of the failed implants were from the maxilla and that too with poorer bone qualities.

Robert A jaffin[73] et al, carried out a five year study on 1054 branemark implants placed in different bone types and reported a 3% failure rates out of which 10% of implant failures were found to be in type 4 bone described by the authors as the ones with thin cortical bone and medullary strength with less dense trabeculae.

Similarly **smedberg et al**[74] conducted clinical trails and reported 36% failure in poor bone types. Hermann et al found that implant failures were strongly correlated to patient factors including bone quality especially when combined with poor quality bone.

This study was conducted in 899 patients as a 3 year follow up by **Bertil Friberg et al, Torsten jemt, ULF Lekholm**[75] and a total of 4641 branemark dental implants were followed up right from stage 1 procedure to the completion of final restoration which revealed only 1.5% failure among them most were the ones in maxilla in which the bone was resorbed and the overall quality of bone was also poor.

Hutton et al[76] conducted a 3 year follow up study involving 133 patients who were implanted with 510 implants of which a combination of implants and overdentures were placed in the maxilla and mandible and the failure rates in maxilla(27.6%) were approximately nine times that of mandibular overdentures(3.3%). And it was inferred that the patients with the least dense bone quality were more likely of failures. And as a whole all of these failures were not principally related to surgical healing but rather happened after prosthetic loading.

Similarly clinical studies were conducted by **Hoar J E[77], Kline R[78], Steiganga[79], Weng[80], Higuchi D[81], Minsk L[82]** and concluded that bone density was a definite factor for prognosis of implant restoration.

Similar FEA studies were carried out by **Mattes danza[83], Tada S[84], Qian L[85], Himmlova L[86], Sutpideler M[87], Cruz M[88]** and concluded that the lower the bone quality, higher the distribution of stress within the bone.

The composite beam analysis[88] is an engineering principle that states when two materials of different elastic modulus are placed together a stress concentration increase will be seen in where the two materials first come in contact similarly when two different entities (such as bone and titanium) with different material properties coexist in a clinical situation the crestal bone and the neck of implant are subjected to this biomechanical imbalance with augmented stress levels thereby bringing about a typical V shaped marginal bone loss as it progresses from the crest to the apex of the implant and the same V shaped stress distribution can be seen in photoelastic and finite element studies when the implant and surrounding structures were loaded with occlusal forces[89]

Bindermann[90] et al in 1970 conducted studies on 54 implant models using two dimensional finite element analysis and concluded that implants tolerated axial loads better than oblique loads and the highest stresses were recorded at the crestal regions of the bone.

In this study the maximum stresses were recorded at the neck of the implant irrespective of the bone qualities and angulation of loads so this area seems to be the weakest link in the design as this is the region where two different materials with different young’s modulus meet. Hence under similar mechanical loading conditions implants generate greater stress ad strains to the overall bone and at the crest of the bone compared with natural teeth[91]

Stress distribution for abutments and implant bodies influenced by loading angulations:

In the present study the von mises stresses on the abutment recorded a mean of 43% difference in stress values between a abutment loaded at 0° and 15°. A 27% difference in stresses for 5° and 10° was observed. And a 22% increase for 10° to 15° with significant P value of (0.001).

The percentage difference recorded for stresses in implant bodies were a mean of 32% difference in stress values between those loaded at 0° and 15°. A 4% for implants loaded at 0 and 5 degrees. A 13% between 5° and 10° loaded implants and a 18% increase for 10° and 15° angle difference.

Papavasiliou et al[7] along with his colleagues conducted studies on IMZ implants through FEA simulations to find out the load values where there will be microfractures he introduced varying conditions such as difference in restorative materials, the bone quality etc. and reported that oblique forces are deleterious with as much as a 15% increase in stress levels and a 200N increase in force increased the levels of stress to about 10 times.

Clelland[93] **NL** et al conducted studies on six implants loaded in three different angulation to analyse stress distribution patterns by photoelastic strain analysis and strain gauges and reports that there was significant increase in stress and strain rates for each increase in angulation of abutments.

Misch et al[98] describes that the magnitude and direction of force are predominatly determined by patient factors such as age, gender arch location, masticatory dynamics, crown height, parafunction. Any oclusal load applied to the implant is divided into,

a. normal force (compressive and tensile)

b. shear force

Not only bone is weakest to shear loads but forces applied at an angle to the bone also further affects the physiologic limit of the compressive and tensile limits of the bone and the angled load typically increases the shear component[99]. A 15° angulation of force increases the buccal component of force by 25.9% and a 30 degree increases it by 50%.

Finite element studies were conducted individually by **Soltesez U**[100]**, Akca K**[101]**, Sertgoz A**[102]**, Ha CY**[103]**, Falk H**[104] on dental implants which simulated a wide array of clinical situations with different loading patterns and reported that non axial loads increased the stresses distributed to the implant and the surrounding bone.

Ko CC et al[105] conducted studies on various simulation models through FEA and elucidated that maximum stresses were on the crest and apex region of the alveolus. And reported about the middle thirds, however in many studies the stresses on interfacial region were difficult to analyse due to the complexity of that area. By utilising the homogenous theory he has also reported microstrain values in the interfacial zone.

Barbier and Shepers[109] conducted studies on implants loaded in dogs which were loaded both axially and non – axially and inferred that greater strains were recorded for the off axis loads.

Rangert B[106] conducted a retrospective study on implant fracture with 39 patients as his study group in whom both anterior and posterior implants were taken into consideration. The study concludes that most of the fractures seemed to take place in which there was cantilevered load to the implants and other overload situations such as bruxism.

A similar retrospective study was conducted by **Kinsel R P**[107] wherein he analysed porcelain failures in implant supported crowns and implant supported fixed partial dentures. And concluded that there was a seven fold increase in failures in bruxers, where there is a occlusal overload due to offset shear forces.

Duyck J[108] he conducted a invivo study on 13 implant supported full prosthesis subjected to forces from various angulations and reported that there was a definite magnification of stresses and that they needed additional implants to bear the additional loads generated.

It is necessary to conduct further studies to analyse stress distribution in order to evaluate fatigue failure, microstrains rates for loading at different angulations, and finally the evaluation of von mises stresses at the unique interfacial zone through other experimental methods such as strain gauges and photoelasticity should be undertaken and finally the gold standards such as prospective longitudinal and multicentre clinical studies are the ones to conform the predictions made from the finite element analysis presented here.

SUMMARY AND CONCLUSION

Implant dentistry is the second oldest discipline in dentistry. The history of implants dates back to ancient civilization several thousand years ago. Ages later, Branemark started experimenting the microcirculation in bone marrow, which lead to the invention of osseointegration subsequently dental implants in the early 1960's. In this study finite element methodology was used to study the stress patterns in seven different zones of which three of them were the implant parts and the rest four were the crown superstructures and the surrounding bone.

All four CAD models were created using CATIA V5 R19 software and then converted into a mesh model by Hypermesh V 13.0 and the final analysis was conducted by ABAQUS 6.14-2 analysis software by loading at vertical axis and offset loading with an angle change brought about in increments of 5° scale with a load of 300 N. The result obtained was further subjected to statistical analysis.

Well within the limits of the study, The following conclusions are drawn

1. The highest recorded von mises stresses were in the neck of the implant region for D1, D2, D3, D4 densities.

2. The von mises stresses for all the seven regions of interest showed highest values (deleterious stresses) for the D4 bone quality.

3. The least stresses compared to other groups were recorded for all 7 regions in the D1 model.

4. The highest stresses were recorded for the D1, D2, D3, D4 bone models when loaded at a 15 degree angulation to the long axis of the implant

5. Likewise, the least stresses among the models loaded in different angulations were for the 0 degree angulation (parallel to the long axis of the implant body).

Conclusion:

In conclusion, the findings of the study indicates,

- In D1 bone quality (dense) the stresses recorded were less and uniform well within the limits of the structures involved whereas in D4 bone quality (less dense) the stresses recorded were of deleterious quality that could bring about deformation.

- The maximum stresses were recorded in the region of neck of the implant for all four bone models.

- The stresses were gradually increased as the angulation of the loading increases from 0° to 15°.

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TABLE 1

SL.NO		D1(0°)	D2(0°)	D3(0°)	D4(0°)
1.	Abutment	283.011 283.033 282.122 283.362	286.363 286.671 286.542 286.172	221.847 221.911 221.432 221.142	221.853 222.331 222.714 221.121
2.	Connecting screw	13.408 13.425 12.422 13.244	12.408 12.777 11.542 12.153	20.612 21.022 21.522 19.932	20.632 21.059 21.546 20.512
3.	Implant	538.938 539.555 539.612 539.282	258.598 258.598 257.123 258.323	776.319 776.319 776.721 776.886	776.338 771.183 770.558 775.432
4.	Co-Cr	379.130 379.152 380.151 379.540	380.085 381.043 382.142 380.448	372.031 372.681 372.127 372.632	372.039 372.064 371.123 372.121
5.	Porcelain	234.700 234.701 234.642 234.612	234.695 234.696 235.512 234.821	234.581 234.684 234.122 234.447	234.581 235.480 235.155 235.742
6.	Cortical bone	64.215 64.159 63.120 64.232	59.163 59.685 58.682 59.721	179.176 179.998 179.332 179.721	179.179 179.648 179.147 178.112
7.	Cancellous bone	1.602 1.663 1.772 1.332	1.392 1.449 1.432 1.822	2.630 3.072 3.357 2.630	2.130 2.226 2.741 2.031
8.	Neck of Implant	539.128 539.121 538.111 538.177	259.121 259.120 259.131 259.322	778.111 778.101 777.110 778.120	777.125 772.121 771.447 777.612

TABLE 2

SL.NO.		D1(5°)	D2(5°)	D3(5°)	D4(5°)
1.	Abutment	286.248 286.312 286.411 286.122	289.203 289.201 289.151 289.134	218.327 218.343 218.121 218.097	218.343 218.444 218.512 217.630
2.	Connecting screw	13.392 13.512 12.128 13.172	12.397 12.412 12.541 11.776	21.540 21.642 21.445 21.711	21.574 21.672 21.443 21.077
3.	Implant	534.463 534.461 538.471 535.512	251.382 251.392 251.488 251.212	861.730 861.742 860.532 861.011	861.772 861732 862.721 861.121
4.	Co-Cr	391.887 391.888 391.772 392.222	392.393 392.411 392.428 391.345	390.413 390.422 390.332 390.211	390.431 389.440 390.344 390.319
5.	Porcelain	225.344 226.351 226.357 225.127	225.338 227.333 226.727 225.820	225.225 226.320 225.676 225.342	225.225 224.220 223.323 225.332
6.	Cortical bone	66.858 67.851 67.721 65.182	62.242 62.250 62.357 62.127	192.129 192.131 191.820 193.521	192.132 192.131 193.111 192.421
7.	Cancellous bone	1.441 1.552 1,551 1.332	1.366 1.145 1.415 1.181	3.056 3.057 2.829 3.711	2.476 2.512 2.741 2.056
8.	Neck of implant	535.140 536.150 535.111 536.568	252.130 253.152 253.552 252.161	862.121 862.132 959.771 862.488	862.135 861.120 862.239 862.576

TABLE 3

Sl.No		D1(10°)	D2(10°)	D3(10°)	D4(10°)
1.	Abutment	387.391 387.401 386.321 387.382	388.357 388.415 387.412 388.435	308.578 308.671 308.152 308.700	308.596 308.612 307.642 308.812
2.	Connecting screw	13.563 13.567 13.511 13.672	13.298 13.722 13.357 13.530	23.711 23.722 24.181 23.701	23.751 22.791 23.55 23.112
3.	Implant	676.151 676.161 675.172 675.566	295.813 295.819 294.152 295.382	958.996 958.997 959.777 957.661	959.044 959.412 959.622 959.157
4.	Co-Cr	499.491 500.452 499.451 497.432	498.016 498.210 498.128 498.442	493.577 493.600 492.521 492.312	493.597 493.611 493.444 491.142
5.	Porcelain	281.285 281.222 281.121 281.444	230.565 230.620 230.447 230.256	230.479 231.412 231.432 231.512	230.479 230.512 231.327 231.308
6.	Cortical bone	75.040 75.412 75.481 75.232	78.026 78.720 78.791 77.621	237.608 237.129 238.322 237.111	237.597 237.642 236.312 236.322
7.	Cancellous bone	1.710 1.712 1.812 1.711	1.634 1.734 1.721 1.711	3.473 3.512 2.829 3.241	2.815 2.817 2.240 2.424
8.	Neck of Implant	677.749 677.749 676.743 677.721	296 296.182 296.543 295.111	959 959.141 959.711 959.512	960 960.141 961.232 961.161

TABLE 4

Sl.No		D1(15°)	D2(15°)	D3(15°)	D4(15°)
1.	Abutment	496.737 496.747 497.322 496.772	499.194 499.232 499.522 499.121	399.154 399.212 399.121 399.344	399.174 399.175 399.111 399.357
2.	Connecting screw	14.321 14.311 13.222 13.322	14.145 14.150 14.111 14.141	26.006 26.412 26.132 26.121	26.051 26.122 26.441 26.298
3.	Implant	813.866 813.877 813.612 813.861	363.026 363.212 362.122 362.343	1181.703 1181.712 1181.332 1181.343	1181.731 1181.732 1182.321 1181.382
4.	Co-Cr	675.097 675.097 674.121 676.772	672.874 672.875 672.821 672.322	651.022 651.212 652.445 652.712	651.038 650.320 651.176 651.151
5.	Porcelain	249.818 250.642 250.643 249.122	249.813 249.815 248.721 249.511	249.730 250.160 250.672 249.614	249.730 249.632 248.172 249.189
6.	Cortical bone	91.030 91.122 90.144 91.382	94.535 94.540 93.322 93.421	295.164 295.177 294.812 295.121	295.150 295.151 295.348 295.151
7	Cancellous bone	2.034 2.420 2.441 2.172	1.937 1.942 1.211 1.128	4.163 4.162 4.774 4.134	3.349 3.352 3.351 3.349
8.	Neck of Implant	814 814.172 815.152 815.572	364 365.121 365.572 363.132	1182 1181.111 1183.126 1183.338	1182 1183.120 1183.111 1183.521